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# How does knee brace modelling influence the prediction of medial and lateral contact forces?

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**Abstract.** Valgus knee orthoses are recommended to improve the quality of life of osteoarthritic patients and to postpone the arthroplasty. An innovative method, based on biplane x-rays imaging and soft tissue stiffnesses, was developed in this study to calculate the forces and moments applied by the orthosis on the skeleton during quasi-static squat postures (at 0°, 15°, 30°, 45°, 70° of knee flexion). This study demonstrated a modification of the adduction-abduction moment and the appearance of a flexion-extension moment applied by the orthosis at high knee flexion. This study also demonstrated a medial compartment unloading when the orthosis was modelled using its 3D movements relative to the bones. High variability between patients was observed. To improve the assessment of orthosis performances, this study suggests to consider the 3D forces and moments applied by the orthosis on the segments of the lower limb.

**Keywords:** Osteoarthritis; Valgus Knee Orthosis; Musculoskeletal modelling.

## 1 Introduction

Knee osteoarthritis (OA) is a chronic disease characterized by joint cartilage damage, with prevalence about 25% for people from 65 years old. Due to severe pain, 80% of OA patients got limited range of motion and 25% cannot practice daily activities anymore ([WHO Scientific Group 2003](#)). To improve the life quality of OA patients and postpone a potential total knee arthroplasty, valgus knee orthoses are prescribed by medical practitioners with the aim of realigning, stabilizing and relieving the joint. It is assumed that the forces and moments applied by the orthosis to femur and tibia/fibula through the soft tissues would help the patients regain an asymptomatic articular motion and loading. These forces and moments applied by the orthosis are of paramount importance when inverse dynamics and musculoskeletal modelling studies are

conducted. Still, the modelling of the mechanical actions of orthosis remains generally simplified to an external abduction moment (Fantini Pagani, Böhle, et al. 2010; Kutzner et al. 2011; Della Croce et al. 2013; Clouthier et al. 2018; Brandon et al. 2019; Hall et al. 2019). To the authors' knowledge, no previous study has modelled the 3 components of the forces and moments applied by the orthosis on the skeleton.

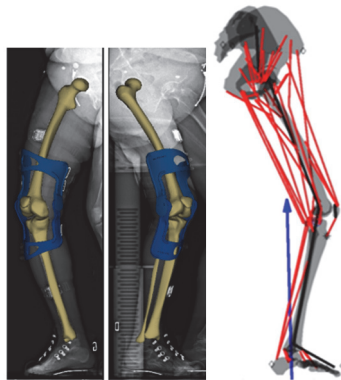
This study aims to (1) develop a model of the orthosis mechanical actions related to the knee flexion angle based on biplane x-rays images and to (2) analyse how customising these forces and moments could modify intersegmental and tibiofemoral contact loads during a quasi-static squat task.

## 2 Material and methods

### 2.1 Experimental measurements

**Participants and protocol.** The experimental data were obtained from 16 patients (5 males and 11 females,  $59 \pm 19$  years,  $93 \pm 49$  kg,  $1.68 \pm 0.09$  m) with severe medial knee OA of Kellgren-Laurence grade 4. All participants completed the consent form approved by the research centre of the CRCHUM and ETS (École de Technologie Supérieure) ethics committees.

A personalised valgus orthosis (Evoke, OssKin, Ortho Inc., Montréal, Canada) was fabricated for each participant using 3D scanning, CAD modelling with a correction of 4 mm in the medial direction and  $5^\circ$  in abduction and 3D printing. A 4-week adaptation period was considered for wearing the orthosis before doing the measurement session. Five weight-bearing quasi-static squat postures were adopted by the patients ( $0^\circ$ ,  $15^\circ$ ,  $30^\circ$ ,  $45^\circ$ , and  $70^\circ$  of knee flexion) for which a pair of EOS low dose biplane x-ray images were acquired (Zeighami et al. 2020) (Fig.1). An AMTI force plate (ORS-6) was fixed inside the EOS cabinet and a platform was designed to isolate the ground reaction forces (GRF) under the contralateral foot, while both feet were maintained at the same level. Measurements were performed twice: braced and unbraced. Two participants have been excluded due to incomplete data collection.



**Fig. 1.** Biplane x-ray images of the weight-bearing quasi-static squat posture at  $15^\circ$  of knee flexion wearing the orthosis and musculoskeletal model.

**Orthosis movements relative to the bones.** The positions of 6 markers on both femoral and tibial parts of the orthosis and the geometries of the bones were reconstructed in 3D from the biplane x-ray images. Coordinate systems were embedded to the 3D bone reconstructions following the standardisation of the International Society of Biomechanics (ISB) (Wu et al. 2002). The displacements and rotations of the orthosis parts relative to the corresponding bones have been computed (Söderkvist and Wedin 1993) between the standing posture ( $0^\circ$ ) and all other squat postures ( $15^\circ$ ,  $30^\circ$ ,  $45^\circ$ , and  $70^\circ$  of knee flexion). Rotations were expressed as a sequence of ZXY Euler angles.

## 2.2 Modelling

**Orthosis mechanical actions on the body segments.** The forces,  $\mathbf{F}_k^i$ , and moments,  $\mathbf{M}_k^i$ , applied by the orthosis on lower limb the segments in posture  $k$  were computed via the measured orthosis movements relative to the bones ( $\mathbf{d}_k^i, \boldsymbol{\theta}_k^i$ ) and the soft tissues stiffness matrices  $\mathbf{K}^i$  of the shank ( $i = 2$ ) and thigh ( $i = 4$ ) segments:

$$\begin{pmatrix} \mathbf{F}_k^i \\ \mathbf{M}_k^i \end{pmatrix} = [\mathbf{K}^i] \begin{pmatrix} \mathbf{d}_k^i - \mathbf{d}_0^i \\ \boldsymbol{\theta}_k^i - \boldsymbol{\theta}_0^i \end{pmatrix} \quad (\text{Eq. 1})$$

The stiffness matrices taken from the literature (Shafiei and Behzadipour 2020) were transported at the knee joint centre (Loncaric 1987).

In the standing posture, the orthosis movements relative to the bones  $\mathbf{d}_0^i, \boldsymbol{\theta}_0^i$  were not measured but have been defined based on the orthosis fabrication principle: tibial orthosis part was translated by  $d_z = 4\text{mm}$  in the medial direction and rotated by  $\theta_x = 5^\circ$  of abduction relative to the femoral part. This way, only lateral-medial displacements and forces and only adduction-abduction angles and moments were assumed to be applied in this standing posture. The mechanical actions on the shank and thigh segments were in opposite directions due to static equilibrium of the orthosis (Eq. 2):

$$[\mathbf{K}^2] \begin{pmatrix} \mathbf{d}_0^2 \\ \boldsymbol{\theta}_0^2 \end{pmatrix} = -[\mathbf{K}^3] \begin{pmatrix} \mathbf{d}_0^4 \\ \boldsymbol{\theta}_0^4 \end{pmatrix} \quad \text{and} \quad \begin{pmatrix} 0 \\ 0 \\ d_z \\ \theta_x \\ 0 \\ 0 \end{pmatrix} = \begin{pmatrix} \mathbf{d}_0^2 \\ \boldsymbol{\theta}_0^2 \end{pmatrix} - \begin{pmatrix} \mathbf{d}_0^4 \\ \boldsymbol{\theta}_0^4 \end{pmatrix} \quad (\text{Eq. 2})$$

To ensure the same static equilibrium in the other squat postures  $k$ , the forces and moments acting along each direction of space kept their sign but their amplitudes were revised (Eq. 3):

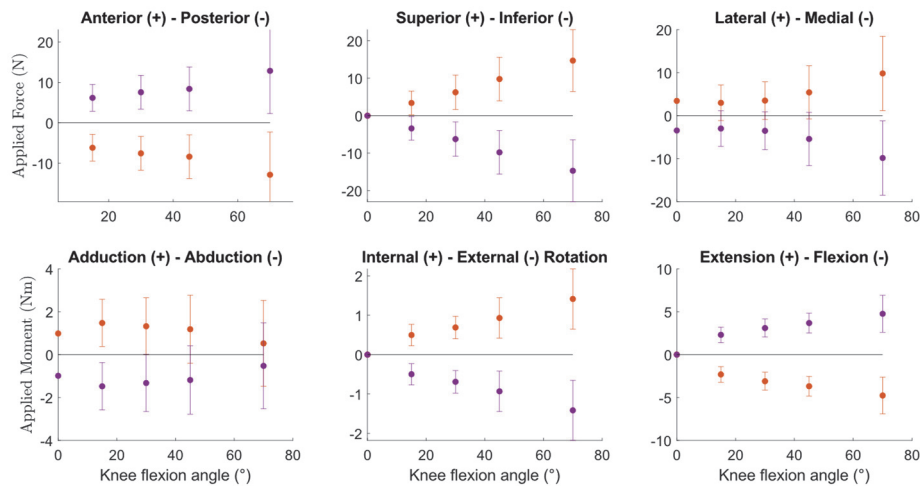
$$\begin{pmatrix} |\mathbf{F}_k^i| \\ |\mathbf{M}_k^i| \end{pmatrix} = \begin{pmatrix} \frac{|\mathbf{F}_k^2 + \mathbf{F}_k^4|}{2} \\ \frac{|\mathbf{M}_k^2 + \mathbf{M}_k^4|}{2} \end{pmatrix} \quad (\text{Eq. 3})$$

**Inverse dynamics and musculoskeletal modelling.** A five-segment model (foot, shank, patella, thigh, and pelvis,  $i = 2$  to 5) was considered with five joints degrees of freedom and forty-three muscles lines of action (Zeighami et al. 2021) (Fig.1). Inverse dynamics and static optimisation (including patient-specific contact point locations) were used to compute intersegmental forces and moments and medial and lateral tibio-femoral contact forces considering the orthosis mechanical actions on the shank and thigh segments: either the forces and moments defined in the standing posture ( $\mathbf{F}_0^i$ ,  $\mathbf{M}_0^i$ ) were assumed constant regardless of squat position (*constant* model) or the forces and moments ( $\mathbf{F}_k^i$ ,  $\mathbf{M}_k^i$ ) were computed from the orthosis movements relative to the bones at each posture  $k$  (*customised* model). Not considering these orthosis mechanical actions was also tested (*no* model). Both intersegmental and applied forces and moments were expressed in the joint coordinate system of the knee (Derrick et al. 2020).

### 3 Results

**Orthosis movements and mechanical actions.** For the orthosis displacements relative to the bones, mainly a displacement in the superior-inferior direction was observed during knee flexion for both femoral and tibial parts. This displacement reached -12.4mm in average at 70° of knee flexion for the tibial parts. The biggest rotation appeared in the sagittal plane (flexion-extension) for the femoral orthosis part (-11.0° in average at 70° of knee flexion). Mean internal-external rotations, of 4.7° and -3.3° for the femoral and tibial parts, respectively, were found at 15° of knee flexion.

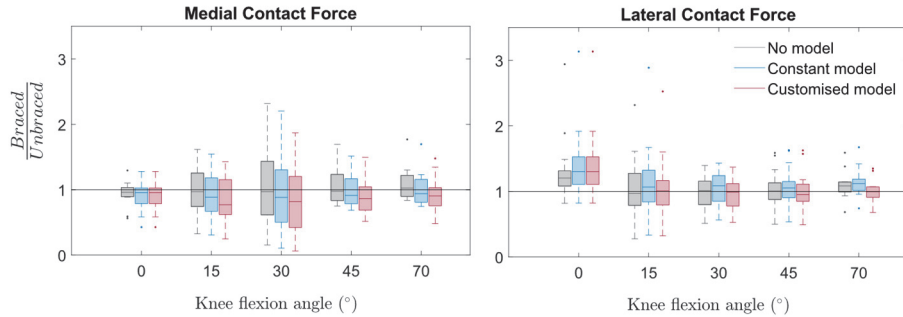
In the standing posture, the medio-lateral forces and the adduction-abduction moments applied by the orthosis on both shank and thigh segments reached 3.4N and 1.0Nm in absolute values (Fig. 2). These loads were the same for all patients and correspond to a unique orthosis design ( $d_z = 4\text{mm}$  and  $\theta_x = 5^\circ$ ). The forces increased with knee flexion and reached 12.8N, 14.7N, and 9.8N in average at 70° of knee flexion in the anterior-posterior, superior-inferior, and lateral-medial directions, respectively. The mean flexion-extension and internal-external rotation moments also increased with knee flexion (up to 4.8Nm and 1.4Nm) while the mean adduction-abduction moment decreased, almost cancelling at 70° of knee flexion. High variability between patients was observed.



**Fig. 2.** Orthosis forces and moments at the knee joint level applied to the shank (purple) and thigh (orange) segments, expressed in the ISB coordinates systems (knee flexion angle should have been defined negative but was reversed to represent the successive squat postures). Dots represent means and lines represent standard deviations at each posture.

**Intersegmental forces and moments and tibiofemoral contact forces.** An increase in the lateral intersegmental force appeared when considering the orthosis mechanical actions applied on the shank and thigh segments. At  $70^\circ$  of knee flexion, the increase reached  $9.8\text{N}$  (*customised* model) when the intersegmental force was  $2.7\text{N}$  (no model). Similarly, the mean abduction intersegmental moment reached, in absolute value,  $5.8\text{Nm}$  in the standing posture (no model) and was reduced by  $1.0\text{Nm}$  (both for *constant* and *customised* model) in all postures. At  $70^\circ$  of knee flexion, the extension intersegmental moment was decreased from  $52.3\text{N.m}$  (no model) to  $47.6\text{N.m}$  (*customised* model).

Ratios between braced and unbraced conditions for the medial contact forces (Fig. 3) appeared mainly modified when the orthosis mechanical actions were considered. A general decrease was observed in the flexed postures with maximal decrease reaching 15% in average at  $15^\circ$  of flexion (*customised* model). In the same time, the lateral contact force appeared slightly increased (*constant* model) or almost unchanged (*customised* model), except for the standing posture with clear augmentation above 20% (all 3 models). High variability between patients was observed.



**Fig. 3.** Ratios between braced and unbraced conditions for the medial and lateral contact forces with and without considering the orthosis mechanical actions on the body segments (no model, *constant* and *customised* models) at each posture (knee flexion angle should have been defined negative but was reversed to represent the successive squat postures).

## 4 Discussion

This study proposed an innovative method to compute 3D forces and moments applied by a patient-specific knee orthosis on lower limb segments based on biplane x-ray images and soft tissue stiffnesses taken from the literature and analysed the intersegmental and tibiofemoral contact loads computed with and without considering the orthosis mechanical action on the body segments.

The measured superior-inferior displacement appeared to be consistent with previously reported orthosis migration (Pierrat et al. 2015). The applied adduction-abduction moments seemed in the lower range of values reported in the literature (between 2.2Nm and 15Nm during gait), when computed from brace frontal plane deflection angles and stiffness, strain gauge or force sensor measurements (Fantini Pagani, Potthast, et al. 2010; Cusin et al. 2018; Brandon et al. 2019). The applied flexion-extension moment at 70° of flexion, revealed to be also lower than the values reported in previous works (McGibbon et al. 2021; Bishop et al. 2023).

Concerning the tibiofemoral contact forces, previous studies have similarly reported a decrease in medial contact forces but with a corresponding increase in lateral contact forces when wearing an orthosis (Pollo et al. 2002; Kutzner et al. 2011; Brandon et al. 2019; Hall et al. 2019). When considering the 3D forces and moments applied by the orthosis, and particularly the flexion-extension moment, the lateral contact forces were not found systematically increased leading to a decrease of the total contact forces in the present study. Indeed, for OA patients, it has been already demonstrated that medial and total contact forces are correlated to both adduction-abduction and flexion-extension intersegmental (or external) moments (Kumar et al. 2013; Zeighami et al. 2021; Amiri et al. 2023; Holder et al. 2023) which, in turn, were found modified in the present study by the moments applied by the orthosis. The greatest alterations were obtained when considering not only the orthosis fabrication principle in the standing posture but also the measured movements of the orthosis relative to the bones (*customised* model).

Some limitations of this work can be noticed. First, each orthosis part was considered as rigid bodies in order to compute their movements relative to the bone segments. Besides, the potential orthosis sliding on the skin patient was not considered. However, this sliding has been reported low compared to the soft tissue deformations (Pierrat et al. 2015). Second, the soft tissue stiffness values were obtained from existing literature and were therefore not patient-specific. Moreover, some studies showed differences in the applied adduction-abduction moment according to the orthosis tightening (Cusin et al. 2018). The computation of orthosis mechanical actions resulting from all these modelling assumptions led to results that can compromise orthosis equilibrium. To address this issue, a control step was incorporated in this study.

## 5 Conclusion

To conclude, intersegmental and tibiofemoral contact loads are dependent on the orthosis modelling approach. In order to enhance the evaluation of orthoses performances, this study suggests to consider the 3D forces and moments applied by the orthosis on lower limb segments. Using the soft tissue stiffness, these orthosis mechanical actions can be defined theoretically based on the orthosis fabrication principle or experimentally based on the movements of the orthosis relative to the bones. In future investigations, using these modelling approaches in a musculoskeletal modelling could enable to study tibiofemoral contact forces for OA patients during dynamic activities such as gait.

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