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► To cite this version:

Sacha Guitteny, Rachid Aissaoui, Ali Zeighami, Raphaël Dumas. Subject-specific osteoarthritic knee joint modelling for contact forces estimation during squat. SB 2022, 47eme Congrès de la Société de Biomécanique, Oct 2022, Monastir, Tunisia. pp S137-S139. <hal-03862291>

HAL Id: hal-03862291

<https://hal.science/hal-03862291v1>

Submitted on 21 Nov 2022

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Subject-specific osteoarthritic knee joint modelling for contact forces estimation during squat

S. Guitteny^a, R. Aissaoui^b, A. Zeighami^b and R. Dumas^a

^aUniv Lyon, Univ Eiffel, Univ Lyon 1, LBMC UMR T_9406, Lyon, France; ^bDépt Génie des systèmes, ETS, Montréal, Canada

1. Introduction

Knee osteoarthritis (OA) is a chronic disease characterized by joint cartilage damage and is a major issue in today's society. Tibio-femoral (TF) contact forces are of crucial importance in the follow up of OA but in vivo measurement of these loads is not feasible.

In response to this issue, load estimation from musculoskeletal models have been widely developed and used since the 60s. Most of the studies that deals with OA patients focused on gait analysis (Meireles et al. 2016; Zeighami et al. 2018) where the patterns of TF contact forces in the medial and lateral compartments have been often reported. Conversely, only few studies have analyzed squat movement and even less with OA patients. Besides, the prediction of TF contact forces with musculoskeletal models in healthy subjects during squat remains inaccurate: the study of (Nejad et al. 2020) reveals a root mean square error of 105.7%BW. Moreover, the distribution of contact forces between knee lateral and medial compartments appears to be different in squat versus gait (Bedo et al. 2020). Different approaches have been suggested in the literature to model the knee joint: hinge, parallel mechanisms or deformable model with contact behavior and ligaments strain. Personalization of the models is often limited to scaling the musculoskeletal geometry and some adaptations of muscle lines of action. Only few models personalized the knee joint (i.e. kinematics, contact points ...).

This study used a musculoskeletal model with subject-specific knee joint and aimed to evaluate its ability in predicting TF contact forces in both medial and lateral compartments during a quasi-static squat task performed by OA patients.

2. Methods

2.1. Experimental set-up

The experimental data used in this study were obtained from 9 patients with severe medial knee OA of Kellgren-Laurence grade 4. All the subjects

completed the consent form approved by the research center of the CHUM (CRCHUM) and ETS (École de Technologie Supérieure) ethics Committees. Five weight-bearing squat postures were adopted by the patients (0°, 15°, 30°, 45°, 70° knee flexion) for which a pair of EOS low dose x-ray biplane images were acquired. Two inertial Sensors (Noraxon, Inc) were placed in the sagittal plane of the shank and the thigh to control the knee flexion/extension angle in real-time. An AMTI force platform (ORS-6) was fixed inside the EOS cabinet to measure the forces and moments under the studied foot. A platform was designed to isolate the ground reaction forces (GRF) under the contralateral foot, while both feet were maintained at the same level.

2.2. 3D bone reconstructions and TF contact points

Subject-specific TF contact points locations (Figure 1) were computed from femur and tibia 3D bone reconstructions (Zeighami et al. 2017).

2.3. Musculoskeletal modelling

A five-segment model (foot, tibia, patella, femur and pelvis), based on (Delp et al. 1990), with five joint degrees of freedom and 43 muscles lines of action was used to represent the lower-limb in this study. All segments were positioned such that reconstructions of bones and anatomical landmarks and the muscles geometry was scaled to the subject anthropometry using the segment lengths. The kinematics constraints of the tibiofemoral joint were described with contact point locations (Zeighami et al. 2018) at each static posture in the EOS cabinet. The tibial insertion of the patellar tendon was located at the tibial tuberosity and the patellar tendon length was fixed as the distance between patellar tendon insertion on the tibia and the distal apex of the patella, from EOS images.

For all the static squat postures, the contact forces and musculo-tendinous forces were calculated via static optimization with a simultaneous minimization (Moissenet et al. 2014).

3. Results and discussion

The predicted medial TF contact forces were obtained between 0.07*BW and 1.08*BW for all squat postures. Only a slight fluctuation of medial forces was observed while the lateral contact forces increased greatly with the knee flexion angle for all subjects. Lateral TF contact forces were in the range

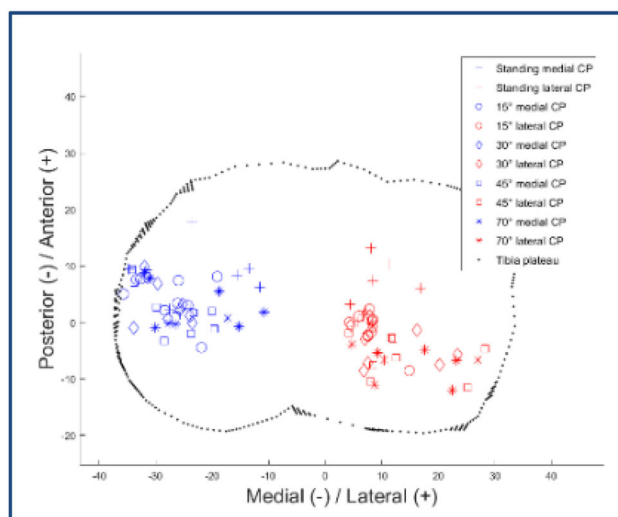


Figure 1. Locations of contact points in tibia plateau at each squat posture.

of $[0.08-0.35 \cdot BW]$ and of $[0.67-1.91 \cdot BW]$ for standing position and deepest flexion, respectively. Thus, most of the subjects' models predicted medial TF contact forces to be higher than the lateral TF forces for small flexion angles (under about 30°) and lower at deeper squat postures. Figure 2 represents the medial (left) and lateral (right) TF contact forces predicted by the subject-specific musculoskeletal models for all patients.

4. Conclusions

This study aimed at evaluating the TF contact forces during a quasi-static squat in an OA group. It used the musculoskeletal model developed by Zeighami et al. (2018) in which the TF joint is personalized. The TF contact forces predicted by our model reveals to be close to the experimental measures of I. Nejad et al. (2020). Indeed, the peak total TF force measured with the instrumented prosthesis was $2.6 \cdot BW$ on average while their musculoskeletal model over predicted an average of $4.59 \cdot BW$. The subject-specific model used in the present study enables to find lower values for the total peak force of $1.92 \cdot BW$ on average. The musculoskeletal model used in the study of Bedo et al. (2020) also reported high values for TF contact forces: peak medial and lateral TF forces were $4.61 \cdot BW$ and $2.4 \cdot BW$, respectively. The lateral contact pattern observed in this study seems in accordance with the curves given by Bedo et al. (2020) during squat. This study has some limitations to be highlighted. First, a positioning support with adjustable height helped the participants to keep the posture in the EOS cabinet and the patients were allowed to

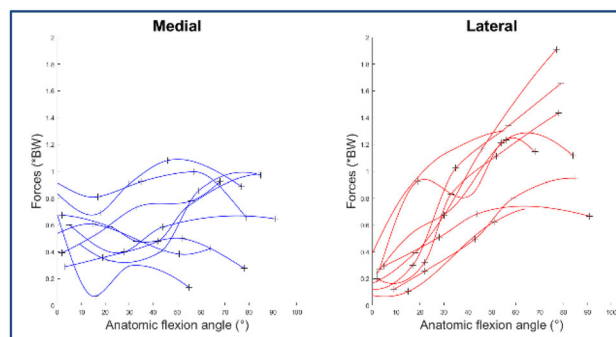


Figure 2. Medial and lateral TF contact forces for each patient for all squat postures. Anatomical flexion angle is between tibia and femur from EOS images.

hold on to the EOS cabinet walls. This was necessary for a major part of the patients that felt pain at deep squat postures. Thus, the measured GRF could be different from a standard squat movement. Second, this movement is also quasi-static. However the study of Clément et al. (2014) reveals only small differences between quasi-static and fast-dynamic squat in terms of 3D kinematics and kinetics. Third, the entire pelvis bone was not visible on the EOS images which prevent from getting an accurate pelvis orientation. For future investigations, the non-linear scaling of tibia and femur bone models can be implemented to personalize the muscular insertions, mostly at the knee level.

To conclude, this study completes the results reported by Zeighami et al. (2017) about altered contact point location in OA patients by computing TF contact forces directly during squat.

Disclosure statement

No potential conflict of interest was reported by the authors.

Funding

This study was partially financed by Région Auvergne Rhône Alpes (PAI 2021), and MedteQ, Québec.

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KEYWORDS Osteoarthritis; tibiofemoral contact forces; musculoskeletal modelling; contact point location

 raphael.dumas@univ-eiffel.fr