

## MUSCULOSKELETAL SQUAT SIMULATION EVALUATION BY MEANS OF AN INSTRUMENTED TOTAL KNEE ARTHROPLASTY

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Knowledge of internal loading conditions based on validated musculoskeletal (MS) simulations can allow improved training and rehabilitation design and monitoring. The aim of this study was to evaluate the ability of individualised reference MS models, based on comprehensive motion analysis, to determine knee joint contact forces (JCFs), as measured in 6 subjects with instrumented total knee arthroplasties (TKA) during squat exercises. Maximum simulated JCFs reached approximately 100% higher than the in vivo measured values at high joint flexion angles; however, at knee flexion angles of below  $\sim 10^\circ$ , the models underestimated the real forces by up to 50%. Improvements of reference MS models, even if they are individualised, are clearly required, especially at joint angles  $< \sim 10^\circ$  or  $> \sim 50^\circ$ .

**KEY WORDS:** SQUATS, LOADING CONDITIONS, INSTRUMENTED IMPLANT, JOINT CONTACT FORCES, MUSCLE FORCES.

**INTRODUCTION:** Knowledge of the internal loading conditions, such as muscle and joint contact forces (JCFs), in strength exercises could provide a strong evidence based foundation for improving training and rehabilitation design and monitoring. While external loading conditions during training exercises such as squats are well known (List et al., 2013; Lorenzetti et al., 2012), internal loading conditions still remain unclear during many exercises, due to the fact that direct, non-invasive access to the muscle and JCFs is not currently available. To access such internal loading conditions, musculoskeletal (MS) simulation plays a key role (Schellenberg et al., 2015), but the accuracy of such analyses, especially if large ranges of motion are required, is known to be sensitive to the subject-specific kinematic, physiological and anatomical specification, which often differs in a complex manner from the modelling data (Correa & Pandy, 2011). The aim of this study was to evaluate the accuracy of simulated JCFs in the knee by means of individualised reference MS models based on the measurement of 6 subjects with an instrumented total knee arthroplasty (TKA) during squat exercises.

**METHODS:** 6 subjects (5m, 1f, aged  $68 \pm 5$  years, mass  $88 \pm 12$  kg, height  $173 \pm 4$  cm) were measured while performing five valid repetitions of a squat exercise without additional weight to avoid overloading (Figure 1). Each subject possessed an INNEX knee implant (Zimmer, Switzerland; type FIXUC), in which the tibial component was instrumented with a 9-channel telemetry transmitter recording with a frequency of 90-100 Hz, which allowed six-component load measurements of the 3 contact forces and 3 contact moments acting on the tibial component (Heinlein et al., 2007). The femur component was unmodified. To analyse the motion of the body, 55 skin markers were attached mainly to the lower extremities (List et al., 2013), and an opto-electronic system (Vicon, Oxford Metrics Group, UK) with 22 cameras (MX40 and MX160) captured the kinematics at a sampling frequency of 100 Hz. The ground reaction forces were measured using two force plates (type 9281B and 9287B Kistler, Switzerland), one under each foot, at a frequency of 2 kHz. All measurement systems recorded simultaneously and were time synchronised. Each subject also performed basic motion tasks according to List et al. (2013) to functionally determine the centres of rotation (CoRs) of the hip, knee and ankle joints. The kinematic and kinetic data were reconstructed in Vicon Nexus (v1.8.5, Oxford Metrics Group, UK) and further processed using Matlab

(R2014a, Mathworks, Natick, MA, USA) to extract skin marker and joint centre locations in space for each time frame, as well as joint angles and ground reaction forces. MS simulation software (OpenSim SimTK 3.3, Stanford, USA) was then used to estimate muscle and tibio-femoral JCFs (Figure 1). Muscle forces were calculated using a static optimization process that minimized the sum of the squared muscle activation at each time frame. Total JCFs were then calculated as the sum of the inverse dynamics resultant forces and all muscle forces that crossed the articulating joint. The definition of the beginning and end of the squat repetition was based on the vertical velocity of the two shoulder skin markers ( $V_{\text{shoulder}} > 20$  mm/s). Segment rotations were computed according to the conventions of the joint coordinate system introduced by Grood and Suntay (1983) to calculate joint angles.

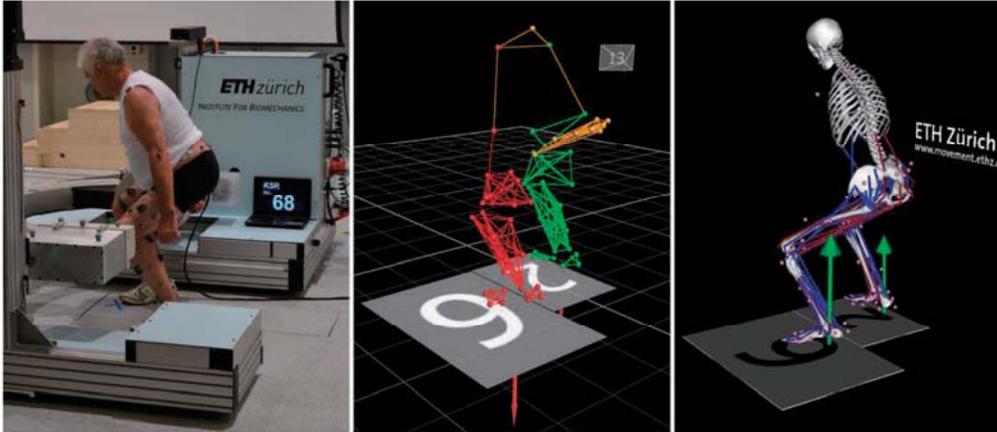


Figure 1: Processing chain from the measurement (left) to the reconstruction in Vicon (middle) and the musculoskeletal simulation in OpenSim (right), including ground reaction and joint contact forces.

Individualised MS models (OpenSim; Delp et al., 2007) were constructed to assess the resultant joint kinetics and kinematics at the ankles, knees and hips of each subject. Here, the reference “Gait2392” model (Anderson & Pandy, 1999, 2001; Delp et al., 1990; Yamaguchi & Zajac, 1989) was adapted to include 14 body segments. The reference model therefore comprised 29 degrees of freedom (DoFs), including 3 DoFs in each knee and ankle joint. Six virtual markers were also included in the model at the CoRs of the hip, knee and ankle joints. The registration of the reference model to the subjects’ specific segment lengths was based on the distances between the functionally defined CoRs (fCoRs) of the hip, knee and ankle as determined from the kinematic data. The fCoRs virtual markers, as well as the skin marker trajectories and the pre-calculated joint angles, were then all used to drive the OpenSim model. Here, the fCoRs for hip, knee and ankle were weighted 100, 100 and 60 respectively. For the skin makers, a soft tissue automated weighting procedure based on Heller et al. (2011) and Krutzenstein et al. (2012) using the relative variance of distance between each skin marker and the corresponding segment centre of mass was used. In order that each segment was considered with equal importance, the sum of all skin marker weighting factors was defined to be 10 on each segment, independent of the number of markers attached to that segment. Similarly, the pre-calculated joint angles were included in the simulation and weighted in all planes (29 DoF; weighting 0.02).

Total JCFs for each subject were calculated for each complete repetition cycle ( $JCF_{MS}$ ) and compared against the forces measured using the instrumented implants ( $JCF_{TKA}$ ). Furthermore, the extreme values were extracted and used to calculate the normalised difference between the simulated and measured JCFs as follows:

$$\Delta JCF^* = \frac{(JCF_{MS} - JCF_{TKA})}{JCF_{TKA}}$$

Additionally,  $\Delta JCF^*$  was calculated as a function of the knee flexion angle and averaged for all repetitions of each subject.

**RESULTS AND DISCUSSION:** Maximum joint contact forces ( $JCF_{TKA,max}$ ) measured on the tibial plateau in the instrumented TKAs during the squat exercises were between 2.0 and 3.3 times bodyweight (BW) (Table 1). Maximal simulated JCFs ( $JCF_{MS,max}$ ) exceeded the measured JCFs by approximately 100%. However, minimum simulated JCFs ( $JCF_{MS,min}$ ), which occurred at low flexion angles during the squats, were lower than the forces measured ( $JCF_{TKA,min}$ ).

Table 1: Mean maximal and minimal joint contact forces (JCF) of the measured knee contact forces (TKA) and of the forces calculated using musculoskeletal simulation (MS) for 5 repetitions of each subject, as well as the average over all 6 subjects.

Subject	Instrumented Implant		MS Simulation		Normalised Difference	
	$JCF_{TKA,max}$ [N/BW]	$JCF_{TKA,min}$ [N/BW]	$JCF_{MS,max}$ [N/BW]	$JCF_{MS,min}$ [N/BW]	$\Delta JCF^*_{max}$ [ ]	$\Delta JCF^*_{min}$ [ ]
1	2.50	1.16	5.08	0.49	1.03	-0.58
2	2.54	0.86	5.92	0.77	1.33	-0.11
3	1.97	0.75	4.63	0.53	1.35	-0.30
4	3.21	0.94	4.72	0.86	0.47	-0.08
5	2.13	0.88	6.70	0.94	2.15	0.07
6	3.25	1.34	5.33	0.88	0.64	-0.34
<b>Average</b>	<b>2.60</b>	<b>0.99</b>	<b>5.40</b>	<b>0.75</b>	<b>1.08</b>	<b>-0.25</b>

Compared to other studies analysing daily activities (Heller et al., 2005) such as gait, running or stair descent, the normalised differences within the present study ( $\Delta JCF^*$ ) were higher, which could be explained due to the higher knee RoMs of squats compared to normal activities of daily living. JCFs were considerably underestimated at flexion angles between  $0^\circ$  and  $10^\circ$ , but greatly overestimated (by a factor of up to 3) the forces at angles above  $\sim 50^\circ$  (Figure 2). Interestingly,  $\Delta JCF^*$  increases almost linearly until a knee flexion angle of approximately  $90^\circ$ . The authors of the reference OpenSim gait model have commented that inaccuracy may occur during high knee flexion angles, which is clearly observed in the present study. Although high loading conditions are known to occur at the position of highest knee flexion (Lorenzetti et al., 2012), are therefore of greatest interest for sports and rehabilitation, these positions are now known to be estimated with least accuracy. As a result, simulated muscle and joint contact forces for activities including high knee flexion angles should be interpreted carefully. Nevertheless, comparison of different strength exercises should still be possible if similar RoMs are used during the different exercises. Despite the fact that the validation of MS models using an instrumented implant is currently the gold standard, there are indeed a number of limitations to this study. It is clear that this study is based on only a small population, performed in elderly subjects, and using implants where the cruciate ligaments are sacrificed. Furthermore, the instrumented TKAs are calibrated using a coordinate system on the tibial component at the height of the PE insert, while the simulated JCFs act at the CoR, possibly resulting in small discrepancies in the comparison of forces.

**CONCLUSION:** In this study, for the first time a reference MS model has been evaluated by means of an instrumented TKA during strength exercises. Interestingly no obvious differences could be observed between the concentric and eccentric phases. The accurate MS simulation of strength exercises seems to be more complex than simulating daily activities due to higher joint RoMs and higher loading conditions. Simulated JCFs in the knee were underestimated at low flexion angles compared to measured JCFs, while in positions of deep knee flexion, simulated JCFs exceeded measured JCFs by up to a factor of 3. Improvements of standard reference models to enable more accurate simulation in high knee flexion angles is clearly required.

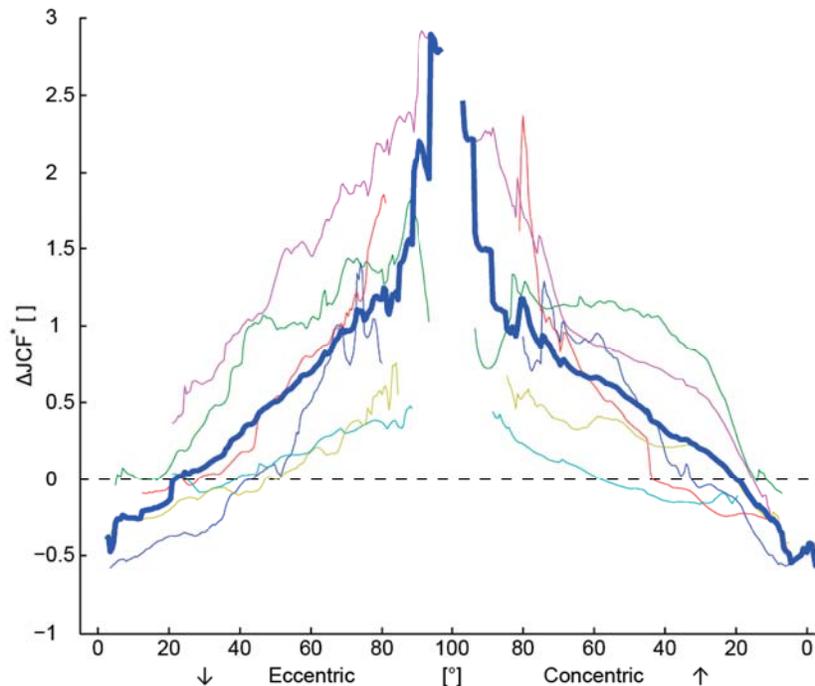


Figure 2: Normalised difference  $\Delta JCF^*$  as a function of the knee flexion angle, starting with the eccentric phase. Mean differences for all repetitions over all 6 subjects are shown as the thick blue line, while the thin lines represent the mean over all repetitions for each individual subject.

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