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Technical note

Evaluation of the accuracy of musculoskeletal simulation during squats by means of instrumented knee prostheses☆

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ABSTRACT

Standard musculoskeletal simulation tools now offer widespread access to internal loading conditions for use in improving rehabilitation concepts or training programmes. However, despite broad reliance on their outcome, the accuracy of such loading estimations, specifically in deep knee flexion, remains generally unknown. The aim of this study was to evaluate the error of tibio-femoral joint contact force (JCF) calculations using musculoskeletal simulation compared to *in vivo* measured JCFs in subjects with instrumented total knee endoprostheses during squat exercises.

Using the early but common “Gait2392_simbody” (OpenSim) scaled musculoskeletal models, tibio-femoral JCFs were calculated in 6 subjects for 5 repetitions of squats. Tibio-femoral JCFs of 0.8–3.2 times bodyweight (BW) were measured. While the musculoskeletal simulations underestimated the measured knee JCFs at low flexion angles, an average error of less than 20% was achieved between approximately 25°–60° knee flexion. With an average error that behaved almost linearly with knee flexion angle, an overestimation of approximately 60% was observed at deep flexion (ca. 80°), with an absolute maximum error of ca. 1.9BW. Our data indicate that loading estimations from early musculoskeletal gait models at both high and low knee joint flexion angles should be interpreted carefully.

1. Introduction

Accurate knowledge of the internal loading conditions in the human musculoskeletal (MS) system, including muscle and joint contact forces (JCFs), can provide a strong evidence-based foundation for improving rehabilitation concepts and customising training programmes, as well as for optimising implant designs. Although direct, non-invasive access to muscle and JCFs is, in most cases, not possible, internal loading conditions have become widely available using MS simulation [1]. The accuracy of such simulations, especially if large ranges of motion (RoMs) are considered, is known to be sensitive to a wide variety of parameters, but often differs in a complex manner from the real *in vivo* situation [2,3].

During strength exercises, extreme postures with large joint angles such as high knee flexion during squats is of interest since the greatest joint loading conditions are thought to occur in these postures [3,4]. While modelling approaches to estimate these internal muscle and JCFs have become widely available in a variety of software modelling packages (e.g., Anybody, OpenSim, LifeModeler^{TM}, MSIM, Biomechanics of Bodies, etc.), validation of the output of MS simulations, especially at higher joint flexion [5], remains difficult due to limited *in vivo* data. One risk is the growing reliance on knowledge gained from standard reference models for application to a wide range of subjects and activities, including rehabilitation and training exercises with high knee flexion angles, without knowing the associated validity of data and the corresponding levels of error. In selected cases, *in vivo* force measurements using instrumented total knee endoprostheses are possible [6], allowing access to the JCFs during a variety of activities, thus providing a reference for evaluating musculoskeletal load analyses. In the recent CAMS-Knee project, such telemetric measurements of *in vivo* knee joint loading were combined with

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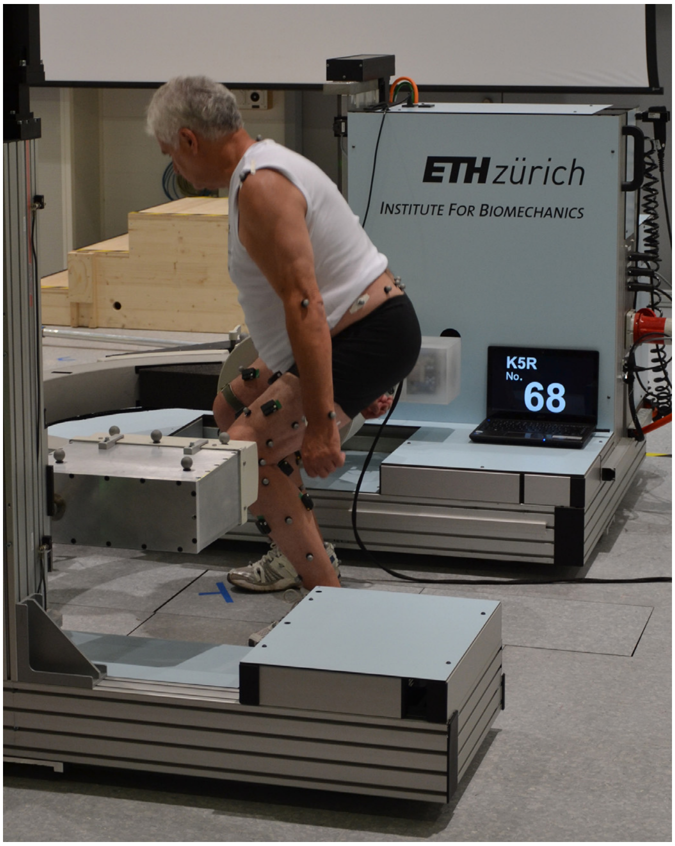

Fig. 1. Experimental set-up. One participant performing a squat within the c-arm of the moving fluoroscope.

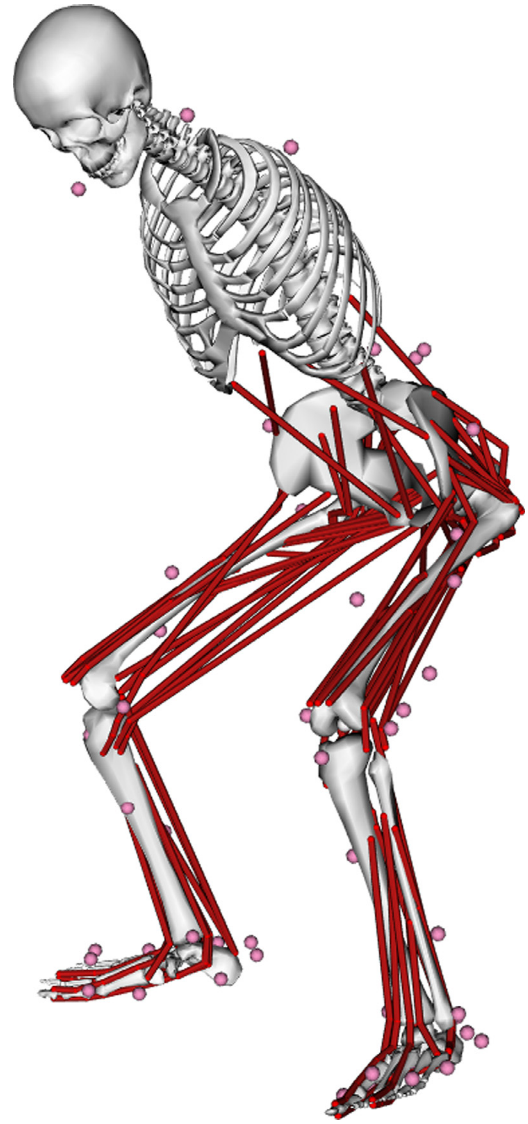

Fig. 2. Musculoskeletal model [OpenSim SimTK 3.3, Stanford, USA; 9] based on the common "Gait2392_simbody" model [10–15] including 14 body segments, 23 degrees of freedom and 92 muscles, shown for an exemplary subject during the squat exercise.

detailed analyses of skeletal motion using moving videofluoroscopy in 6 subjects [7]. Using these unique data, the aim of this study was to evaluate the accuracy of JCFs determined using open source MS simulation tools compared to tibio-femoral JCFs measured in 6 subjects each with an instrumented total knee arthroplasty (TKA) during squat exercises.

2. Methods

This kinematics and kinetics data used in this study were based on the CAMS-Knee (www.cams-knee.orthoload.com) datasets. While the measurement protocols are described in the literature [7], a brief description of the data is provided here: six subjects (5 m, 1f, aged 68 ± 5 years, mass 88 ± 12 kg, height 173 ± 4 cm) were measured while performing five repetitions of a squat exercise without additional weight. Each subject possessed an INNEX knee replacement (Zimmer, Switzerland; type FIXUC), in which the tibial component was instrumented with a 9-channel telemetry transmitter (90–100 Hz) that allowed six-component load measurements of the 3 JCFs and 3 joint moments acting on the tibial component to be recorded within an accuracy of 2% [6]. To assess the motion of the body, 55 skin markers were attached mainly to the lower extremities [8], and an opto-electronic system (Vicon, Oxford Metrics Group, UK) with 22 cameras (MX40 and MX160) captured the kinematics (100 Hz). The ground reaction forces (GRFs) were measured using two force plates (Kistler, Switzerland; 2 kHz), one under each foot. The CAMS-Knee study was approved by the local ethics committees of the Charité (EA4/069/06) and ETH Zürich (EK 2013-N-90) and all subjects provided written informed consent prior to participation.

Each subject performed basic motion tasks to functionally determine the centres (fCoRs) and axes of rotation of the hip, knee and ankle joints [8]. The squat activity then consisted of each subject standing with stationary feet, approximately shoulder width apart, starting with the eccentric phase (Fig. 1). Knee joint flexion was then performed as far as possible before returning to the standing position. The kinematic and kinetic data were reconstructed in Vicon Nexus (v1.8.5, Oxford Metrics Group, UK) and further processed using in-house software written using Matlab (R2014a, Mathworks, USA) to extract skin marker and joint centre locations for each time frame, as well as joint angles and GRFs.

To estimate tibio-femoral JCFs, MS simulations of each subject's lower limbs were created that included scaling to their segment lengths (determined from the fCoRs), inverse kinematics and inverse dynamics to determine the joint moments, and finally optimization to determine the muscle forces, as described below. The models [OpenSim SimTK 3.3, Stanford, USA; 9] were based on the common "Gait2392_simbody" model [10–15] including 14 body segments, 23 degrees of freedom and 92 muscles [16] (Fig. 2) without residual reduction analysis [17]. Each model's segment lengths, as well as muscles' slack and tendon lengths, were scaled based on the fCoRs of the ankle, knee and hip [9]. To reconstruct the

Table 1
Extreme (max or min) joint contact forces (JCF) of the knee measured in subjects using instrumented total knee arthroplasty (TKA) or calculated by means of musculoskeletal simulation (MS) as well as the Averaged Peak Error [%].

Subject	Instrumented implant		MS simulation		Averaged Peak Error	
	$JCF_{TKA,\min}$ [BW]	$JCF_{TKA,\max}$ [BW]	$JCF_{MS,\min}$ [BW]	$JCF_{MS,\max}$ [BW]	$Error\ JCF_{min}$ [%]	$Error\ JCF_{max}$ [%]
1	1.16	2.50	0.53	3.95	−54.1	58.2
2	0.86	2.54	0.44	4.20	−48.9	65.5
3	0.75	1.97	0.59	3.32	−21.0	68.6
4	0.94	3.21	0.87	5.11	−7.5	59.0
5	0.88	2.13	0.47	3.42	−47.3	60.7
6	1.34	3.25	0.59	4.52	−55.8	39.2
Ave.	**0.99**	**2.60**	**0.58**	**4.09**	**−39.1**	**58.5**
Std	**0.20**	**0.49**	**0.14**	**0.62**	**18.2**	**9.4**

subjects' kinematics, marker weightings were chosen as follows: hip, knee and ankle with 50; skin markers with 1, and an automated weighting procedure based on soft tissue artefacts, with a total weight of 10 for each segment [18–20].

To calculate muscle moment arms, especially for multi-joint muscles, the OpenSim generalized force method was used, which takes wrapping and via points of muscles into account [21]. A static optimization criteria that minimized the sum of the squared Hill-type muscle [16,22] activation was used [23–25]. Total simulated JCFs (JCF_{MS}) were calculated as the sum of the muscle, ground reaction, segment inertial forces, and masses.

To evaluate the accuracy of the quasi-static optimization approach for determining internal forces at the knee, JCF_{MS} were compared against the measured JCFs from the instrumented TKAs (JCF_{TKA}). The *Error JCF* was calculated for each trial as follows:

$$Error\ JCF\ [\text{in}\,\%] = \frac{(JCF_{MS} - JCF_{TKA})}{JCF_{TKA}} \cdot 100 \tag{1}$$

Additionally, the results obtained for all 6 subjects were arithmetically averaged inter-individually and presented as a function of the knee flexion angle [°]. Furthermore, the maximal and minimal measured ($JCF_{TKA,max}$, $JCF_{TKA,min}$) and calculated ($JCF_{MS,max}$, $JCF_{MS,min}$) forces within each cycle were extracted and used to calculate the average peak error ($Error\ JCF_{max}$) between the simulated and measured JCFs.

3. Results

JCFs on the tibial plateau (JCF_{TKA}) of between 0.8 and 3.2 times bodyweight (BW) were measured *in vivo* across the six patients during the squat activity (Table 1 and Fig. 3). The simulated forces for the same activities were calculated as 0.4–5.1 BW, both under- and over-estimating the measured JCF_{TKA} at different flexion angles. The averaged peak error ($Error\ JCF_{max}$) over all subjects and squat cycles predicted by the MS models was 58.5%, while the inter-subject standard deviations of this error were around 9% (Fig. 4). The associated RMS error of the simulated joint centres and marker positions were 10 mm and 13 mm respectively. The average total model residual force (at the pelvis) was 19.6 N.

An almost linear dependence of the *Error JCF* was observed with knee flexion, resulting in single cycle absolute errors of up to 100% at deep knee flexion angles (Fig. 4). The range of knee flexion where the errors in the JCF remained between ± 20% was approximately 25°–60° during both eccentric and concentric movements.

4. Discussion

The usage of reference models for the determination of internal loading conditions has become commonplace, but the accuracy of such gait models, particularly during other more challenging activities that include higher RoMs has, until now, remained

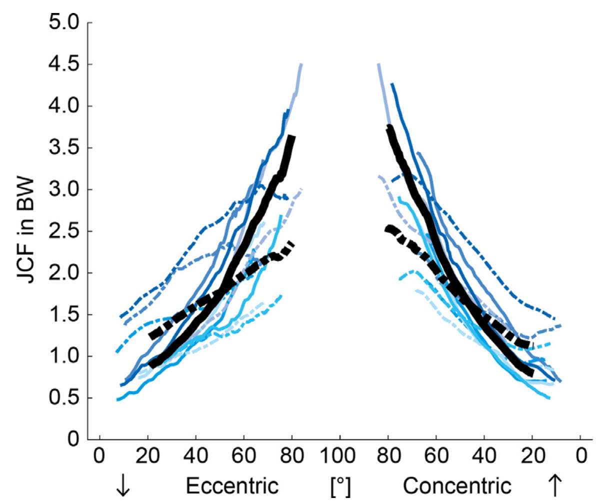

Fig. 3. Joint contact forces (in BW) as measured using the instrumented implant are presented for each subject (thin dashed coloured lines) together with the estimated JCFs from the MS models for each subject (continuous lines; same colour as the measured forces) over the complete range of flexion throughout the squat activity. The mean JCF is shown as a thick line.

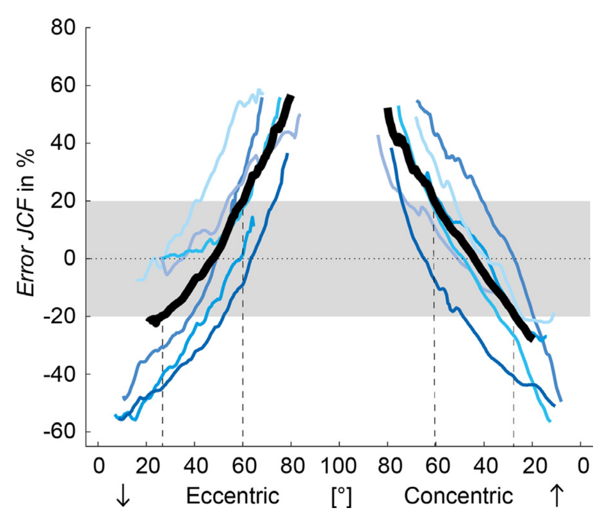

Fig. 4. Errors in the joint contact force (error JCF, shown in %) of the simulated forces are shown compared to the telemetrically measured knee joint contact force across the range of knee flexion achieved during the squat activity. Since not all participants reached the same range of knee flexion, data from only the region where 4 or more trials were available is presented. The thin lines represent the average error in each individual subject, shown in subject specific colours consistent with Fig. 3. The black thick line depicts the average of all 6 subjects (averaged JCF error). The grey areas show the average range of joint flexion (dashed lines; 26.5°–60° eccentric; 27.5°–60.5° concentric) that achieved an averaged JCF error of within ± 20%.

Table 2
Muscle moment arms [cm] around the hip and knee joints for different muscles (m. gluteus maximus, hamstrings, quadriceps, and m. gastrocnemius) at the time points where maximal JCF-errors (under- and over-estimated) occur. Note that some of the lever arms could be considered to be extremely small, thus providing a subjective indication of possible sources of JCF error. Note that the moment arms of all muscles are larger at the time point of JCF underestimation than overestimation.

		Moment Arms at max over-estimation [cm]	Moment Arms at max under-estimation [cm]
Hip	glutMax1	-1.9 ± 0.5	-5.3 ± 0.7
	glutMax2	-2.4 ± 0.6	-5.9 ± 0.7
	glutMax3	-4.1 ± 0.9	-6.6 ± 0.2
	glutMed1	0.0 ± 0.4	-2.7 ± 0.8
	glutMed3	-2.0 ± 0.3	-3.4 ± 0.6
	glutMin1	-0.2 ± 0.3	-1.6 ± 0.7
	glutMin3	-0.7 ± 0.2	-2.2 ± 0.7
	biFemLh	-4.5 ± 0.9	-5.9 ± 0.6
	semimem	-4.2 ± 0.8	-5.4 ± 0.5
	semiten	-5.4 ± 0.9	-6.3 ± 0.8
	rectFem	4.4 ± 0.8	4.7 ± 0.6
Knee	biFemLh	-1.9 ± 0.4	-3.4 ± 0.1
	biFemSh	-2.4 ± 0.4	-3.1 ± 0.2
	semimem	-3.1 ± 0.4	-3.6 ± 0.2
	semiten	-3.6 ± 0.5	-4.4 ± 0.3
	rectFem	3.0 ± 0.3	5.2 ± 0.2
	vasInt	3.0 ± 0.2	4.9 ± 0.2
	vasLat	3.1 ± 0.2	4.7 ± 0.2
	vasMed	3.2 ± 0.2	4.7 ± 0.2
	latGas	-1.4 ± 0.2	-4.2 ± 0.3
	medGas	-1.6 ± 0.1	-4.0 ± 0.3

limited. In this study, a reference open source MS model has been bench-marked against *in vivo* measured forces for six subjects throughout repetitions of the squat activity. Our data suggest a clear relationship between the error in force calculation and the knee flexion angle. However, with an average error range of $\pm 20\%$ (individual errors of up to $+53\%$ or -45%), the results of this study do indicate that the investigated MS model is indeed able to estimate knee JCFs during activities where the knee joint flexion angle remains within approximately 25–60° These data are in agreement with previously observed JCFs during the late stance phase of gait (25%−100% stance phase) but differ from the first 25%, where an overestimation of the maximal JCF has been reported [2,26]. In our study, the lack of impact, shear forces, and one-legged stance phases during squats, which are all present during gait, are likely to play key roles in explaining the observed underestimation of JCFs, and this is consistent with known problems associated with transferability between different activities [27,28]. As a result, adaptation of the results during squatting to other (weight-bearing) activities might be limited and should be handled with care. At high flexion, the lever arms of some muscles were observed to be particularly low (Table 2), possibly resulting in an unfavourable redistribution of the forces throughout other muscles, and well might contribute the large overestimation of JCFs. A large inter-subject variation was also observed (Table 1, Figs. 3 and 4), signifying that results of a single individual should be interpreted with care.

During strength training, the highest loading conditions are associated with positive adaptation but also possible injury, and often occur in positions of deep knee flexion [4]. However, the loading conditions in these postures are now known to be estimated with least accuracy and should therefore be interpreted with caution. The flexion dependent error also possibly explains the larger average peak errors ($Error\,JCF_{max}$) found in our study (58.5%) compared to a previous study analysing daily activities such as gait (11%) or stair climbing (26%) [5], where their activities were performed at lower knee flexion angles. However, since a low peak tibio-femoral JCF error of $14 \pm 10\%$ was observed in their study at ~90° flexion and peak tibio-femoral contact forces of 2.2–2.3BW were simulated by others [29] during squatting activities, it seems that the error is likely to be governed by more than just differences in joint flexion angles alone.

While it is entirely possible that a regression relationship (Error JCF [%] $= 1.3 \cdot x + 57.5$; where x represents the knee flexion angle; $R^2 = 0.99$), could be used to correct predictions when using the Gait2392_simbody model, it should be recognised that systematic improvements to the underlying model should be targeted if possible, rather than adjusting the output forces to reduce errors by means of a correction factor. The clear question posed by the results of our study is: why does the investigated standard MS model have less capacity to predict forces at extreme ranges of knee flexion? Based on the nature of the squat movement, the external knee joint moment increases almost linearly with the flexion angle [4]. This means that the demand on the extensor muscles increases with flexion, suggesting that outside of the ~25°–60° range, the muscle loading is either excessively low or high. It is therefore probable that an improvement to subject specific anatomical parameters [26], dynamic wrapping paths and lever arms of muscles [particularly at the knee and hip; [30,31], as well as the consideration of passive soft- and connective-tissue forces (e.g., ligaments, joint capsule etc.), muscle co-contraction [29], muscle and tendon slack lengths [32], and goal-oriented muscle optimisation [26,33], will all play key roles for achieving improved predictions at high and low flexion angles. In this study, only a subjective assessment of the sources of error could be performed (Table 2), but further investigations to better identify the true origins of the errors in an objective manner should systematically examine the relative importance of each parameter using e.g., Monte Carlo analyses.

One of the strengths of this study was the usage of the complete CAMS-Knee datasets [7] as a basis for the musculoskeletal analyses performed. These datasets combine state-of-the-art measurements of internal joint contact forces at the knee [6,34] with skeletal motion of the knee joint using moving videofluoroscopy [35] and whole-body kinematics using motion capture [36], and will be made available for public release at https://cams-knee.orthoload.com. As a result, the data driving the analysed models is the current gold standard. However, there are a number of limitations to this study to be considered. It is clear that our results are based only on a small population of elderly subjects, using implants where the cruciate ligaments have been sacrificed [3,34]. In this respect, the lever arm of the patella tendon in particular, which is known to play a critical role for the prediction of tibio-femoral JCFs [26], might well differ from the standard, and relatively old version of the Opensim MS models used in our analyses due to the implanted patella button. However, it is important to emphasise that more up-to-date MS models are now available. The next stages of this work will therefore establish whether more detailed models are able to reliably predict JCFs.

In this study, a knee joint flexion dependent error of $< 20\%$ was observed between approximately 25°–60°, suggesting that widely available MS models can allow access to loading predictions at mid-range flexion angles, but indicates that loading estimations should be interpreted carefully. An improved understanding of the aetiology of the potentially large errors is clearly required.

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Conflicts of interest

There are no conflicts of interest.

Funding

We would like to acknowledge the financial support of the Robert Mathys Foundation (RMS) for enabling the measurement of the subjects with telemetric knee implants (within the original CAMS-Knee study: http://cams-knee.orthoload.com/) to occur at ETH Zürich, Switzerland.

Ethical approval

The data used in this study were captured from subjects measured in a previous investigation: the CAMS-Knee project, which was approved by the local ethics committees of the Charité (EA4/069/06) and ETH Zürich (EK 2013-N-90). All subjects provided written informed consent prior to participation.

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