INTRODUCTION

Although total knee arthroplasty (TKA) treatment for knee patients is successful and reproducible, difficulties or pain during motion still persist in a limited number of patients. This might be explained by surgical errors or excessive deviations from the standard knee anatomy, which can lead to a different biomechanical behaviour, than what the prosthesis was designed for. The aim of this work is to estimate and compare the contact forces in four different TKA designs during a loaded deep squat simulating surgical errors and patient-related anatomical factors.

METHODS

CT images of one cadaveric full leg (Caucasian male) were used to generate 3D models of the bones and to obtain a physiological knee model assuming standard positions of the main soft tissue insertions, as described in literature. Four different TKA types were chosen in this study: a fixed bearing, posterior stabilized knee; a high flexion fixed bearing guided motion knee; a mobile bearing knee and a hinge knee. All prostheses were the same size and replaced both cruciate ligaments and all resurfaced the patella. Following the surgical procedure of each TKA, the proper surgical cuts on the bone model were identified and performed. Each TKA was virtually implanted according to the cut bone geometries, thus defining the reference replaced knee model. Each derivative replaced knee model was then obtained by changing the values of one parameters in a range based on literature and surgical experience [1-3]. The following configurations were analyzed:

1. the reference configuration;
2. the change in location over ± 5mm of both proximal and distal insertion points simultaneously of Medial Collateral Ligament (MCL) and Lateral Collateral Ligament (LCL) in medio-lateral (ML), antero-posterior (AP) and proximo-distal (PD) directions to simulate the effect of nonstandard anatomy or of ligament release;
3. the change in location over ± 5mm of the distal patellar tendon insertion in PD direction to simulate the effect of different patellar tendon lengths;
4. the change in position of the tibial component in ML and AP direction over ± 3mm;
5. the change in orientation of the tibial component in flexion-extension (FE) and abduction-adduction (AA) over ± 3° and in internal-external (IE) orientation over ± 5°;
6. the change of the patella in height, simulating patella alta (with Blackburne-Peel index (BPI) of 1.29) and patella baja (with BPI = 0.59) [4,5];
7. the change in orientation of the patellar component in IE orientation over ± 10° simulating patellar tilting.

A loaded squat to 120° was performed for each configuration, with a constant vertical hip load of 200N. These settings match the experimental tests performed in a previous in-vitro analysis on cadaver legs [5-8]. Each replaced model was developed and analyzed using a commercial musculo-skeletal modelling software (LifeMOD/KneeSIM 2008.1.0, LifeModeler Inc., San Clemente, CA) [6,9]. The loaded squat was reproduced numerically, simulating an existing knee kinematics rig [8,10] in terms of geometries, constraints, inputs and outputs (Figure 1). For each model, both the maximum PF and TF contact force have been evaluated. All contact forces are expressed in body weight (BW), referring to a BW of 712 N.

RESULTS AND DISCUSSION

For each design, the PF contact force increased with flexion. It reached a maximum just before contact between the quadriceps tendon and the femoral trochlea occurred, after which it decreased. The patellar peak force with a range of 2.1-6.5 BW, is in agreement with the range reported in literature [11-13].

Table 1 reports the maximum total PF and TF contact force (in the lateral and medial compartment) during the squat for all the TKAs for all the configurations. A patella baja always reduces the maximum PF contact force (up to 32%) while a patella alta always increases the maximum values (up to 67%). An anterior translation of the baseplate increases the maximum contact force (up to 30%), while a posterior translation reduces the contact force (up to 26.7%). An internal patellar tilting can decrease the contact force (up to 29% for one type) while an external tilting increases the PF forces (up to 16.7%). All these results are in agreement with literature [6, 14-17]. The
other tibial component positions and the other configurations affect the contact forces less. For all designs, the medial and lateral TF contact forces increased with flexion. As for the PF contact forces, because of differences in design of the TKAs, the implants show different magnitudes of the maximum lateral and medial tibio-femoral contact forces. Nevertheless, all of them demonstrate similar trends. Like PF forces, also TF maximum forces are high during a squat even in the physiological knee [18,19]. Our results for peak TF force are 1.6-5.4 BW.

The maximum TF contact forces increase for patella alta up to 22% but don’t decrease substantially for patella baja. For one TKA type an internal tilting of the patella induces an increase of the force while for another type it produces a reduction of the force. Looking at antero-posterior translations, medio-lateral translations, slopes and internal-external rotations of the tibial component, the results show that each type has a different behaviour without a general trend. Usually, the changes in TF forces are less than those in the PF force in the same configurations. Medio-lateral, proximo-distal and posterior LCL translations did not induce changes in the peak TF force. Instead, an anterior translation of the LCL usually increases the peak TF force, in both condyles. MCL translations can alter the TF forces but without a general trend among prosthesis types.

CONCLUSIONS
The sensitivity analysis showed that, generally, patellar contact forces are mostly affected by implant position, especially by patellar height and tibial components position, while tibio-femoral contact forces are mostly affected by the displacement of the MCL and by an anterior translation of the LCL. Moreover, although some of the investigated parameters induce similar changes in different TKA types, some other parameters can induce completely different behaviour depending on implant type. Looking at the obtained results, we can conclude that for the definition of a TKA numerical model the selection of the correct boundary conditions (in terms of implants and insertion point positions) is a very critical issue. Much attention should be given to the determination of the initial conditions of the simulation because even a small change in them could influence the resulting contact forces significantly.

Table 1: Maximum patello-femoral contact force and maximum tibio-femoral contact force, in the medial and the lateral compartment, for the analyzed configurations and TKA designs

<table>
<thead>
<tr>
<th>Configuration</th>
<th>PF Medial</th>
<th>PF Lateral</th>
<th>TF Medial</th>
<th>TF Lateral</th>
</tr>
</thead>
<tbody>
<tr>
<td>Patellar Component</td>
<td>A</td>
<td>1.5</td>
<td>1.7</td>
<td>4.7</td>
</tr>
<tr>
<td></td>
<td>B</td>
<td>1.5</td>
<td>1.7</td>
<td>4.7</td>
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<tr>
<td></td>
<td>C</td>
<td>1.5</td>
<td>1.7</td>
<td>4.7</td>
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<tr>
<td></td>
<td>D</td>
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<tr>
<td></td>
<td>E</td>
<td>1.5</td>
<td>1.7</td>
<td>4.7</td>
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REFERENCE