THE INFLUENCE OF INCREASED DOF IN THE KNEE JOINT ON MUSCLE ACTIVATION TIMINGS AND FORCES IN A MUSCULOSKELETAL MODEL

1,2Paulien Roos, 3Ilse Jonkers, Kate Button1,2 and 1,2Robert van Deursen

1School of Healthcare Studies, Cardiff University, Cardiff, UK
2Arthritis Research UK Biomechanics and Bioengineering Centre, Cardiff, UK
3Katholieke Universiteit Leuven, Leuven, Belgium
email: RoosPE@cardiff.ac.uk, web: http://www.cardiff.ac.uk/arcbbc

INTRODUCTION

Altered kinematics following a knee injury can lead to increased knee joint loading [1], which may lead to early onset osteoarthritis [2]. Muscle activation patterns influence contact forces in the knee [3]; muscle forces also play an important role in stabilizing the knee joint [3]. Knee injuries often result in altered muscle activation patterns but it is not known how exactly this influences knee loading. Musculoskeletal models allow investigation of the influence of altered muscle activation patterns on forces acting on the knee. Simulations with a full body model with a simplified knee joint can estimate muscle forces, which can be used as input in complex knee models to investigate detailed loading of the condyles [3]. This pilot study investigated the influence of increased knee joint degrees of freedom (DOF) on the accuracy of muscle onset and kinematic estimations. This will provide insight into the role of active control in frontal plane knee stability and will allow future investigation of the effect of altered muscle activations due to injury.

METHODS

One healthy subject (female, age 31 years, height 1.74 m, mass 59 kg) stood with feet shoulder width apart and body weight distributed over both legs (DS), body weight was slowly transferred to the left leg lifting the right leg slightly off the ground (TR1), this was held (ST) and then shifted back to both legs (TR2). This movement involved minimum knee flexion and allowed focus on the frontal plane. Kinematic data were collected at 250 Hz using a VICON system (Oxford Metrics Group Ltd., UK), and the ‘Plug-in-Gait’ marker set. Ground reaction force data were measured for each foot at 1,000 Hz using two force plates (Kistler Instruments Ltd., Switzerland). EMG data were collected at 1,000 Hz using a Noraxon Telemyo system. Electrodes were placed on the left leg on rectus femoris (RF), vastus lateralis (VL), vastus medialis (VM), tensor fascia latae (TFL), biceps femoris (BF), semimembranosus (SM), lateral (LG), and medial gastrocnemius (MG). Raw EMG was band-pass filtered (20-450 Hz) with a 4th order Butterworth, then full wave rectified, low-pass filtered (10 Hz) with a 4th order Butterworth, and normalized to maximum muscle activation of the simulations. Medio-lateral coactivation (COACTMed-Lat) was calculated using the summed activation of the lateral muscles (VL, TFL, BF, LG) and of the medial muscles (VM, SM, MG) and applying the algorithm proposed by Winter [5]. Simulations were performed in OpenSim (v2.4.0, SimTK, USA). Marker positions and ground reaction force data were transformed and imported. The gait2392 Simbody model [4] was scaled using a calibration trial. Inverse kinematics calculations were performed, followed by Residual Reduction Analysis. Muscle activations and forces were computed using Computed Muscle Control. Muscle activations were low-pass filtered (10 Hz). The gait2392 model has a 1-DOF (sliding hinge joint) knee. A 2nd varus/valgus DOF was added and a simulation was performed using the same experimental data as with the 1-DOF knee model.

RESULTS AND DISCUSSION

Accuracy of the inverse kinematics was the same in the 1-DOF and 2-DOF knee model (squared error: 0.002 m; marker error RMS: 0.009 m). Kinematics differed minimally with maximum RMS differences in joint angles between the models of 0.6°. Adding a varus/valgus DOF influenced muscle activation patterns and forces, and how well muscle activation patterns agreed with EMG (Fig. 1, Table 1).
Peak differences in muscle force changed up to 236 N (table 1). In the DS and TR1 phases VM and BF activity and forces increased while SM activity and forces decreased in the 2-DOF compared to the 1-DOF knee simulation. This asymmetrical activation pattern in the upper leg at the anterior medial and posterior lateral side relates to control of varus/valgus movement of the knee. In the ST phase VL, TFL, BF and LG activity and forces increased, while MG decreased in the 2-DOF compared to the 1-DOF knee simulation. This increased lateral activation in the stance leg would be required to control the center of mass (which is medial of the center of pressure) and keep the right leg lifted off the ground. TR2 showed a similar pattern to DS and TR1 with increased VM, LG and BF activation and forces and decreased SM and MG in the 2-DOF compared to the 1-DOF knee simulation. This asymmetric activation pattern also relates to control of varus/valgus movement. Adding the varus/valgus DOF resulted in decreased COACT\textsubscript{Med-Lat} due to the increased activation of lateral muscles (Fig. 2). In the simulations muscle activations were optimized to minimize control and stability was not taken into account. Knee injured patients may have altered

activation. Alternative optimization algorithms are needed to account for stability and altered control in patients. The model also did not account for medio-lateral stability provided by tibio-femoral contact and ligaments.

CONCLUSIONS

This study showed that adding varus/valgus DOF to the knee influenced muscle activation patterns and forces in simulations with a musculoskeletal model. Including a varus/valgus DOF resulted in decreased medio-lateral coactivation. This may indicate a need for alternative optimization algorithms that account for stability, or a need for a more comprehensive knee model. Further development of these methods is required before active control of the knee in the frontal plane and patient’s muscle activity can be effectively explored through simulation.

REFERENCES


ACKNOWLEDGEMENTS

Rebecca Hemming for her help with data collection. Dr. Roos is an Academic Fellow funded by Arthritis Research UK (Grant No 18461).

### Table 1: Peak difference in muscle force between the simulations with the 1-DOF and 2-DOF knee as absolute value in N and normalized as percentage of the peak muscle force.

<table>
<thead>
<tr>
<th>Difference</th>
<th>RF</th>
<th>VL</th>
<th>VM</th>
<th>TFL</th>
<th>BF</th>
<th>SM</th>
<th>LG</th>
<th>MG</th>
</tr>
</thead>
<tbody>
<tr>
<td>N</td>
<td>123</td>
<td>-96</td>
<td>-118</td>
<td>-40</td>
<td>-126</td>
<td>107</td>
<td>-145</td>
<td>236</td>
</tr>
<tr>
<td>%</td>
<td>27%</td>
<td>46%</td>
<td>49%</td>
<td>44%</td>
<td>50%</td>
<td>43%</td>
<td>58%</td>
<td>56%</td>
</tr>
</tbody>
</table>