1 Musculoskeletal Modelling of the Lower Limb

1.1 Introduction

Knowledge of the internal loads acting on the human body during daily life movements has a wide range of applications, from clinical assessment of motor control patterns to prosthesis design and preclinical testing and as an input for finite element models predicting bone adaptation (Geraldes and Phillips, 2010; Phillips, 2010).

External forces are generally measurable directly through force plates or load cells, while muscle forces and joint reaction forces remain unknown. In the last decade, in vivo joint contact forces acting at the hip (Bergmann et al., 2001) and at the knee (D’Lima et al., 2007; Kutzner et al., 2010) have been recorded by instrumented prostheses, but the results are available only for a relatively small set of patients. On the other hand, direct muscle force measurement is not possible in humans as it is considered too invasive.

In order to provide estimation of these internal forces, musculoskeletal models have been introduced (see Erdemir et al. (2007) for a comprehensive review). These models are composed of rigid bodies representing the bones connected by mechanical joints. Muscles are included
in the model as actuators able to contract and provide the joint torques necessary to accelerate the segments, so generating the body movement.

1.2 The London Lower Limb Model

A musculoskeletal model of the lower limb has been developed in OpenSim (Delp et al., 2007) from the anatomical dataset published by Klein Horsman et al. (Klein Horsman et al., 2007) with the purpose of estimating muscle and joint contact forces during daily living activities. The unilateral model is composed of six bodies (pelvis, femur, patella, tibia, hind foot, mid foot plus phalanxes) connected by a spherical joint (hip), and 4 hinges (patello-femoral joint, tibio-femoral joint, talocrural joint and subtalar joint). Accounting for the 6 degrees of freedom of the pelvis with respect to the ground and for the kinematic constraint used to define the patellar movement, the model has twelve degrees of freedom. A total number of 38 muscles are represented by 163 actuators (Klein Horsman et al., 2007). Viapoints and wrapping surfaces were introduced to improve the path of some muscles. Minor modifications were applied to the original database in order to achieve ankle joint functionality.

Full details of the model are given in:


The developed musculoskeletal model is available from the SimTK London Lower Limb Model website.

The static indeterminacy, due to the number of muscle actuators spanning a joint being higher than the joint degrees of freedom to equilibrate, is overcome through the static optimization technique. This method yields a unique solution to the distribution problem (Crowninshield and
Brand, 1978) by minimizing a function of the muscle activities under the constraint of moment equilibrium for the joints. The muscle forces are also limited to assume a value in a range between zero and the muscle maximum isometric force.

1.3 Validation for daily living activities

According to Morlock et al. (Morlock et al., 2001) the most frequently performed daily living activities are level walking and stair climbing. The lower limb model was therefore validated for these two activities by estimating the hip contact forces and comparing these values with direct measurements obtained from the literature (Bergmann et al., 2001). A cycle to cycle validation was performed, as kinematics, kinetics and hip joint contact forces were available from the publicly available HIP98 dataset, available from the Orthoload website. A comparison with the results of previous validation studies (Heller et al., 2001; Stansfield et al., 2003) was also possible.

The obtained calculated joint forces were generally in close accordance with the measured experimental hip contact forces for both activities.
The contact force predictions provided the best match to the experimental values when minimising a quadratic objective function (Figures 3 and 4).

The parameters used to quantify the accordance were the percentage error at the frame of experimental peak, the relative deviation between peaks, the shift between peaks. As a general indicator of the prediction, the root mean square error and the Pearson’s product-moment correlation coefficient were also calculated. Table 1 summarises the average results.

A more detailed analysis of the model predictions (Modenese and Phillips, 2011) investigated the potential of the model to predict fine adjustments in the muscle recruitment observed when walking at different walking speeds.
Figure 3: Comparison between experimental (red line) and predicted hip contact forces (black line) during level walking for four different patients. Experimental data published by Bergmann et al., 2001.

Figure 4: Comparison between experimental (red line) and predicted hip contact forces (black line) during stair climbing for four different patients. Experimental data published by Bergmann et al., 2001.

<table>
<thead>
<tr>
<th>Activity</th>
<th>Error (peak to peak)</th>
<th>Error (at experimental peak frame)</th>
<th>Time shift</th>
<th>RMSE range</th>
<th>R</th>
</tr>
</thead>
<tbody>
<tr>
<td>Level walking</td>
<td>10.1</td>
<td>20.8</td>
<td>5.9</td>
<td>23.2-52.4</td>
<td>0.90-0.96</td>
</tr>
<tr>
<td>Stair climbing</td>
<td>-7.8</td>
<td>10.0</td>
<td>3.8</td>
<td>20.0-61.1</td>
<td>0.84-0.97</td>
</tr>
</tbody>
</table>

Table 1: Summary of comparison between predicted and measured hip joint contact forces. Experimental data published by Bergmann et al., 2001.
Differences in the representation of the gluteal muscles between (left to right) Arnold et al., 2010, the Klein Horsman dataset, and the LHDL dataset (Testi et al., 2010).

1.4 Future development

1.4.1 Musculoskeletal model

In the performed simulations, the lower limb model tended to overestimate the mediolateral component of the hip joint load. The main reason for this has been identified as the poor geometrical representation of the gluteal muscles, whose structure cannot be adequately represented by a simple straight line approach without leading to a moment arm underestimation (see Figure 5). Therefore further work is currently in progress in order to improve the geometrical representation of the gluteal muscles, especially to preserve the layered structure usually compromised or strongly modified in musculoskeletal models.
1.4.2 Finite element application

The bone geometries of the specimen dissected by Klein Horsman et al. were not made available, so a preliminary registration to different bone geometries is necessary before undertaking the development of finite element models utilising muscle and joint contact forces extracted from the musculoskeletal model. This operation, together with the creation of an OpenSim plugin to directly extract the muscle lines of action from the leg model, is currently in progress for the femur.

1.5 Downloads

The following files related to this project are available for download from the SimTK project page.

- London Lower Limb Model R1.0

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As further releases of the London Lower Limb Model are developed and published these will be made available using a Creative Commons license through the Structural Biomechanics and SimTK websites.
1.6 References


• Morlock, M., Schneider, E., Bluhm, A., Vollmer, M., Bergmann, G., Müller, V. and Honl, M., 2001. /Duration and frequency of every day activities in total hip patients/. Journal of Biomechanics 34, 873-881.


Contact: Luca Modenese (l.modenese08@imperial.ac.uk)  
Structural Biomechanics (structural.biomechanics@imperial.ac.uk)