OPTIMIZATION-BASED MUSCLE FORCE SCALING FOR SUBJECT SPECIFIC MAXIMUM ISOMETRIC TORQUE ESTIMATION

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SUMMARY
This study aimed at developing a force scaling method based on inverse dynamics and static optimization using a small amount of experimental data. The factor K, used for estimating maximum isometric muscle force in musculoskeletal models, was optimized by minimizing the least mean square distance between experimental and simulated joint torque. The muscular force sharing was performed locally for each joint. The method was tested using the data collected from a volunteer subject on an ergometer for hip and knee flexion and extension. Promising results were found for the hip. For the knee, probably more parameters have to be included for defining a personalized model. Limitations of the joint-by-joint muscle force estimation approach were also discussed.

INTRODUCTION
A personalized human model is ideally required for ergonomic analysis of a task performed by an operator (e.g. for identifying main risk factors). Current digital human models can be easily be scaled in terms of anthropometric dimensions. But how to define a subject specific model in terms of one’s physical capacity is still highly challenging. Many researches were performed to determine muscle forces in the past [5], but only few studies have been focused on force-scaling, in order to estimate maximum isometric joint torque and to characterize subject specific physical capacity. Among many parameters of a musculoskeletal model that could be defined as subject specific, according to Scovil and al. [7], one of the most important is the maximum isometric muscle force. For each muscle, this maximal force is proportional to its physiological cross-sectional area (PCSA). So the proportional factor, commonly named K, could be used as a subject specific parameter. The present study aimed at demonstrating the feasibility of a muscle force-scaling for hip and knee joints using this proportional factor K.

METHODS
Musculoskeletal model
A 3D musculoskeletal model [6] was used to estimate lower limb muscle forces. Muscular geometry was taken from the lower limb model developed by S. Delp [2]. Muscles were represented by their line of action going from origin to insertion point. Via points were included when necessary and multiple lines of actions were used to model broad muscles (such as gluteus muscles). For each muscle, the maximum isometric muscle force was considered. Muscles PCSAs were taken from data reported by Thorpe and al [8]. Force-length and force-velocity relationships, tendons, as well as excitation-contraction dynamics were not considered. The muscle forces were estimated for a joint position using a static optimization procedure. It consists of minimizing the squared muscle stresses while satisfying the equality between the sums of individual muscular moments to the joint torques of all DOF:

\[ \min \sum_{i=1}^{n} \left( \frac{F_i(t)}{PCSA_i} \right)^2 \quad s.t. \quad \sum_{i=1}^{n} \overline{L}_A(t) \cdot F_i(t) = \overline{T}_{\text{joint}}(t) \]

The individual muscular moments were estimated from the muscle forces and the muscle lever arms which were derived from musculoskeletal geometry. Performance of this model to reproduce maximal voluntary force was assessed by comparison to literature data [1].

Force scaling method
In this study, one specific factor was estimated to describe subject’s muscular capacity in flexion/extension for a given joint. From maximum voluntary joint torque collected experimentally, the goal of the optimization is to find the factor K that minimizes the difference between experimental and model-based data. The optimization loop consists in two steps.

Step 1 is the determination, for a given joint position, of the muscle force sharing (set of muscle activation \( \alpha_i \)) that 1) produced the maximum joint torque on the axis of interest (e.g. flexion/extension axis); 2) kept joint torques on the other axes at null value. For each posture, the maximum joint torque can be expressed as a function of the parameter \( K_{\text{optim}} \):

\[ \overline{T}_{\text{joint}}^{\text{Max}} = K_{\text{optim}} \sum_{i=1}^{n} \overline{L}_A \cdot PCSA_i \cdot \alpha_i \]

Step 2 is, then, the estimation of \( K_{\text{optim}} \) that minimizes the least mean square distance between experimental and simulated data considering all trials for a given joint.

Experimental data
Hip and knee extension/flexion maximum isometric joint torques were performed by a healthy volunteer subject on a specifically designed ergometer (Figure 1). Each joint was tested in several positions: knee flexion of 15° and 45°, hip
flexion of 45° and 60°. The trials were tested randomly. For each position, the subject was asked to exert a maximum voluntary joint torque and to maintain it during 5s. Each trial was repeated once. The positions of retro-reflective markers attached to the subject body were measured at 100Hz using a 10 cameras video-based motion capture system (VICON, Oxford, UK).

A 7 DOF kinematical model of the lower limb composed of pelvis, thigh shank and foot segments connected by hip, knee and ankle, was then used to reconstruct the subject’s posture for each trial. The joint moments were then computed using a 3D inverse dynamics method based on homogeneous matrices [3, 4].

The proposed method was applied separately to the hip and knee. The muscular force sharing was performed locally for each joint, meaning that bi-articular muscles were considered as two independent single-joint muscles. For each joint, eight trials were considered: four in flexion and four in extension.

RESULTS AND DISCUSSION
At first, joint torques on the axes different from flexion/extension were found very small for both hip and knee joints from experimental data. This agreed with the hypothesis made for muscular force sharing.

Then, for the hip joint, the ratio of simulated torque with optimized factor \( K_{\text{opt}} \) on the experimental torque varied from 0.8 to 1.2, implying an error between -20Nm to +13Nm (Table 1). For the knee joint, the ratio varied from 0.5 to 1.5 with an error of -22Nm to +23Nm. In addition, most of the simulated torques were underestimated. The estimated value of \( K_{\text{opt}} \) was in the range of reported values for the hip, but among the lowest for the knee.

In the current study, only one parameter \( K \) was determined for each joint. Probably, more \( K \) have to be introduced considering muscular functional groups, such as the quadriceps femoris or the hamstrings. A possible solution could be to associate a \( K \) for each functional group.

A joint-by-joint approach was used to estimate muscle forces sharing. This might explain at least partly the results obtained for the knee. Indeed, according to Fraysse and al [5], joint-by-joint approach favors the single-joint muscles whereas the global approach tends to favor bi-articular muscles. This may lead a different muscle force sharing for the knee joint as the most important muscles such as rectus femoris for knee flexion/extension are biarticular. Different muscle sharing would also affect the estimation of \( K_{\text{opt}} \). For the hip, most powerful muscles such as the gluteus maximus for extension are mono-articular. The difference between joint-by-joint and global methods should be much smaller for the hip than for the knee.

CONCLUSIONS
This study proposed a method of personalization of muscular capacities, based on the optimization of the isometric maximum muscular capacity proportional factor \( K \). The results showed promising results on the hip joint, but limitations of the method for the knee. Probably more parameters have to be included for different functional muscle groups. This may also be due to the joint-by-joint approach used for muscle force estimation which underestimates the recruitment of bi-articular muscles. Further investigation is required to verify this idea and to determine the minimum number of the parameters to be introduced for defining a personalized model.

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REFERENCES

Table 1: Results of personalization on hip and joint data

<table>
<thead>
<tr>
<th>( K_{\text{opt}} ) (N.cm²)</th>
<th>( T_{\text{Sim}}/T_{\text{Exp}} ) Mean</th>
<th>Min</th>
<th>Max</th>
<th>( T_{\text{Sim}} - T_{\text{Exp}} ) (N.m) Mean</th>
<th>Min</th>
<th>Max</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip</td>
<td>30.8</td>
<td>1</td>
<td>0.8</td>
<td>1.2</td>
<td>0</td>
<td>-19.8</td>
</tr>
<tr>
<td>Knee</td>
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<td>0.8</td>
<td>0.5</td>
<td>1.5</td>
<td>-6.8</td>
<td>-21.7</td>
</tr>
</tbody>
</table>

Figure 1: Experimental ergometer for hip flexion (a) and knee flexion (b) joint torque measurements