Biomechanical model of the lower limb based on relevant actions for the control of knee-rehabilitation parallel robots

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Parallel robots (PR) are increasingly being used for lower limb rehabilitation due to their robustness, simplicity, versatility, load capacity and low cost. In the last decade, a few rehabilitation parallel robots (RPRs) were developed, mainly for the ankle [1] and, more recently, for knee rehabilitation [2]. Unlike exoskeletons where mechanical actions are applied on the joint axis, RPRs exert mechanical actions over the distal end of the limb. For this reason, the control actions monitored from the robot may be unrelated with those transmitted to the muscles and ligaments. This limits the effectiveness of the exercises as well as the possibility of developing dynamic safety systems. As a result, developing a biomechanical model for the estimation of the relevant forces in the knee, such as muscle, tendon, ligament, and tibial contact forces, while doing the exercise with the robot, will be of a great challenge and interest. Nevertheless, most of the efforts in this area were focused fundamentally on models with exoskeletons [3]. And a little effort was provided in the development of complete models for the lower limb with PR [4]. Moreover, most of musculoskeletal models suffers from its complexity, along with the need of solving the redundancy problem of muscular actions. Also, the models conclude to systems difficult to personalized and very inefficient from a computational point of view [3]. As a result, their practical implementation in rehabilitation tasks is limited because they require estimating internal actions in real time.

In this work a biomechanical model for the control of a PR for knee rehabilitation is presented and validated. This model is subject adaptable and is pretended to work in real time. It is based on the development of the model presented in [5] which is a parametric model that can be personalized from the anatomical measurements of the lower limb and the estimation of the hip joint center. The knee motion is modeled as a four-bar mechanism, which is fitted for each subject by using offline optimization processes. As mentioned in [5], muscle forces are obtained from the inverse dynamic model using an optimization process to solve the redundancy problem. To improve its computational effectiveness, muscle coefficients were estimated offline using a simulated model. It was found that two generalized coordinates were sufficient to span all the extents of the workspace of the pretended rehabilitation task; one for the hip joint and another for the knee. The workspace was discretized using these parameters, then muscle coefficients were calculated accordingly and the values that were stored offline. In the actual real-time robot controlled rehabilitation task, muscle coefficients were retrieved directly according to the values of the generalized coordinates. Figure 1 shows an example for the coefficients of the Rectus femoris muscle coefficients.

In order to check consistencies, the model was used to predict muscle forces during a commonly used knee rehabilitation exercise, namely squat exercise. A healthy person accomplished various cycles of squat exercise subjected to three load levels; without loads (L0), wearing a 6 kg backpack (L1), and wearing a 12 kg backpack (L3). The predicted muscle forces were compared with the corresponding EMG signals for the most important muscles controlling the knee motion. i.e. (gastrocnemius, biceps femoris, and vastus medialis and lateralis muscles). The comparison was made at two levels; i) Correlation between EMG signal amplitude (rms value of raw signal) and the magnitude of the calculated force; and, ii) Predictive capability of the model to estimate the muscle forces in the L1 level based on the calibration made between the EMG signal and the muscle forces at levels L0 and L2.

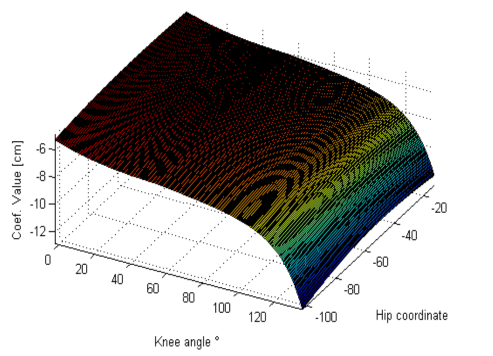

Fig. 1: Rectus femoris muscle coefficients versus the relevant generalized coordinates for the lower limb.

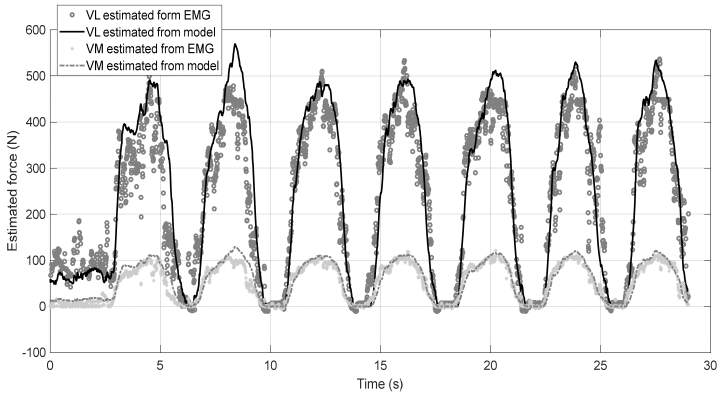

Fig. 2: VL and VM muscle forces, load level L1 (6 kg).

Experimental results show a high correlation between the rms EMG signal amplitude and the magnitude of the force calculated using the model. Table 1 shows the results for vastus lateralis muscle. On the other hand, the calibration made between the rms EMG signal and the muscle forces at levels L0 and L2 was substantially capable to estimate the muscle force in the L1 condition, as shown in figure 2.

Tab. 1: Correlation coefficients between the model results and EMG signal. Vastus laterais muscle.

MOVEMENT	L0	L1	L2
FLEXION	0.960	0.957	0.951
EXTENSION	0.946	0.968	0.946

The results of the validation process show that the proposed model can be used effectively for controlling and monitoring the relevant actions during rehabilitation exercises of knee joint. Offline functional calibration of the hip and knee joint parameters and model-anatomical scaling capability provide a good model personalization. In addition, obtaining muscle coefficients in an offline process greatly minimizes the computational time for real-time control strategies without affecting model precision.

Acknowledgments

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