# SOLVING INVERSE PROBLEMS IN BIOMECHANICS USING LEG EXTENSIONS AS AN EXAMPLE

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Precise analysis of the muscle forces responsible for specific human movements represents a basic task in biomechanics. Here the focus is on gaining detailed information of the interplay between muscle forces and individual sports movements using inverse dynamics.

**KEY WORDS:** biomechanical modelling, inverse dynamics, knee model.

**INTRODUCTION:** The quantification of not directly-measurable attributes for muscle contractions in sport movements may lead to further information for the strength training (Hatze 1998, Wank 2000). The goal of the study was to develop a leg model for the diagnosis of muscle forces in sports movements using inverse dynamics.

**METHODS:** According to the inverse problem of the dynamics of multi body systems (MBS), leg extensions in a leg press are analyzed using a MBS model for the musculoskeletal system of the lower extremities. With this model, internal forces, generated by muscles and tendons, which produce this specific movement, are determined using the known time histories of the internal coordinates and the external forces. Consequently, the anatomical characteristics of muscles, tendons and joints have to be taken into account, as well as energetic characteristics and neurological control tasks.
To estimate the specific movements, the Langrangean equations of motion were integrated for the MBS. Therefore the MBS Modeller Studio alaska 4.2[1] was used. The external forces were measured using dynamometry. The time histories of the analyzed movements were gained by means of video analysis. From these data the inverse kinematics were developed, i.e. the time histories for the spatial coordinates of the man model DYNAMICUS were calculated as well as the appropriate velocities and accelerations using the Dynamic Tracking method (Roemer et al. 2001). The next step was to reproduce the reference movement of the model using Lyaponuv-stable dynamic control (Maißer & Jungnickel 1998) and estimate the joint reactions. Lever rules were applied to calculate the one-dimensional force of the replacement muscles from the joint moments. For the knee joint, an individually parameterized model was used which takes the moving joint axis (MD) of the femoral-tibial part as well as the femoral-patellar part into account (Fig.1). The input data for this model were extracted from MRI scans in 11 different knee positions.

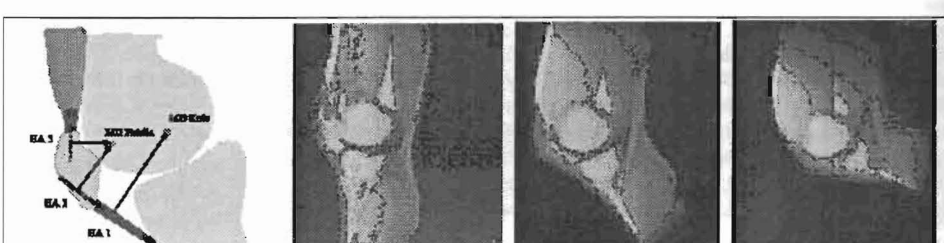
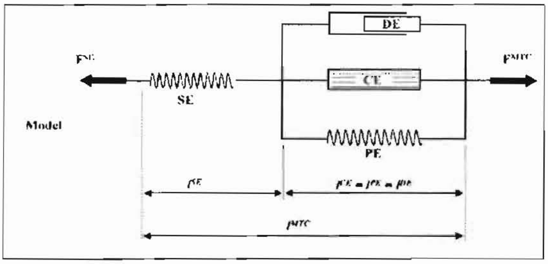


Figure 1: Knee model and MRI scans.

[1] *alaska* Modeller Studio© 4.2, Institute of Mechatronics at the Chemnitz University of Technology

Because of the redundancy of the biological system, the calculations of the muscle forces were simplified by modelling the extending muscles as one single-joint replacement muscle for the thigh (Mm. Quadriceps) and the shank (Mm. Triceps) respectively.
To determine the internal forces, a Hill-type muscle model (Fig. 2) was used with a single serial part, i.e. serial elastic (SE) and three parallel parts, i.e. parallel elastic (PE), damping element (DE), and contractile element (CE).




Further results of the leg model were length and velocity of the replacement muscles from consideration of their insertion points. To determine the lengths and velocities of the single parts within the muscle model we used

$$l^{SE} = \sqrt{\frac{F^{CE} + F^{PE} + F^{DE}}{k_{SE}}} + l_0^{SE} = \sqrt{\frac{F^{MTC}}{k_{SE}}} + l_0^{SE} \qquad (1)$$

$K_{SE}$ was determined by the assumption that

$$k_{SE} = \frac{F_{iso}^{SE}}{\left(1.04\, l_0^{SE} - l_0^{SE}\right)^2} \qquad (2)$$

(Van Soest 1992).

Knowing the length and therefore the velocities of SE, CE, PE and DE, the forces for PE and DE could be calculated. The behaviour of PE was calculated using a quadratic approach and that of DE by a linear approach. Henceforth, the force of CE is the sum of SE, DE and PE.
Beside the determination of the actual forces generated within the replacement muscle, the greatest possible force (F_poss) was calculated for this muscle depending on muscle length and contraction velocity (this model has been described in full detail in Van Soest 1992). This is necessary for a first estimation of whether the results may show an appropriate solution.
Based on F_poss, the muscle activation was determined using the activation function of HATZE (1977, 1981). Q depends on the stimulation, muscle length and time and ranges between 0 and 1.

$$F^{CE} = Q(q\,,\, l^{CE},\, t) \cdot F_{poss} \qquad (3)$$

The input parameters for F_poss were Hill's parameters a and b and the optimal muscle length. Both parameters were optimized so that they fit the measured data. For the measurements single joint tests were used. Therefore the muscles were tested using different loads to measure the force-velocity relationship as well as isometric tests at different joint angles to quantify the force-length relationship.
For the leg press study we used a leg press equipped with two force plates in the foot pad and an odometer and velocimeter applied to the sled.
The time histories of the angles for the metatarsophalangeal joint, the ankle, the knee joint and the hip joint were detected using videometry. The muscular activity was monitored by a 5-channel electromyogram.
Starting position was a pre-set position for the feet on the foot pad with a knee angle of 95°.

The following measurements were taken:

- Quantification of the isometric maximum force in the starting position
- Leg extension with a load of 90% (L_90)
- Leg extension with a load of 70% (L_70)
- Squat jump (SJ).

To measure the ground reaction force (GRF) under isometric conditions the sled was fixed to the bar. For the concentric experiments the external load was excluded from this GRF. For the squat jump, the load was calculated in a way that the participant had to move the same load as he would do in a standing position without additional weights.

**RESULTS AND DISCUSSION:** The load for the isometric maximum was 216 kg. Consequently, the additional weight was 194 kg for L_ 90% and 144 kg for L_70%. To execute a squat jump with a load similar to body weight for a vertical movement, an additional weight of 57 kg was necessary. In the following text, the results for the muscle model will be presented.
In separately conducted single joint tests, we found a maximum isometric force of 6633 N for the contractile element of the Mm. Triceps. Within the leg-press the maximum force for L_70 was only 295 N; for L_90, 646 N; and for SJ, 450 N. These data reveal that the function of the Mm. Triceps in this movement is primarily the stabilisation of the ankle. For SJ the highest value was generated after leaving the foot pad while the ankle moment was already decreasing. Consequently one could assume that the Mm. Triceps is mainly used for the active plantar flexion proceeding after take-off but has only small effects on the jump height for this measured movement.
The Mm. Quadriceps produced a maximum force of 5442 N for L_70. This value corresponds to 80.9% of the isometric maximum of 6721 N detected in single joint tests. For L_90, the internal load was with 6433 N at 95.7%. According to F_poss, the proportion of the internal load increased to 81.4% for L_70 and 96.3% for L_90, respectively.

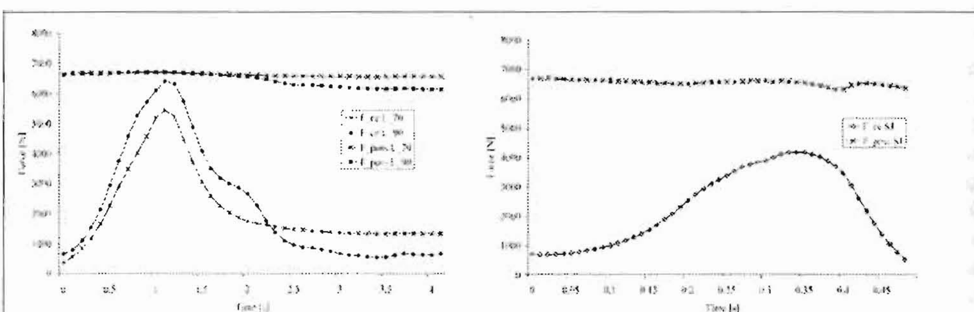


Figure 3: Force time histories of CE and F_poss.

The force for SJ was significantly smaller (4184 N). Despite the faster movement, the higher contraction velocity has only slight effects on F_poss, which was affected to a greater extent by the muscle length. Because of the relatively slow movements, the influence of the damping element on the generated muscle force was of no relevance to the results. The highest maximum force over all tests was 12 N shortly before take off at SJ. The PE generated no detectable force. The reason was that, in the starting position, the muscle length was already shorter than the optimal muscle length. Consequently, there was no stretching during the movement.
Fig. 4 shows the measured and calculated normalized activation functions. The measured activation recorded by electromyography is equivalent to the sum of all muscle parts within the Mm. Quadriceps and Mm. Triceps. The results shown were obtained solely using inverse dynamics without further optimizations.

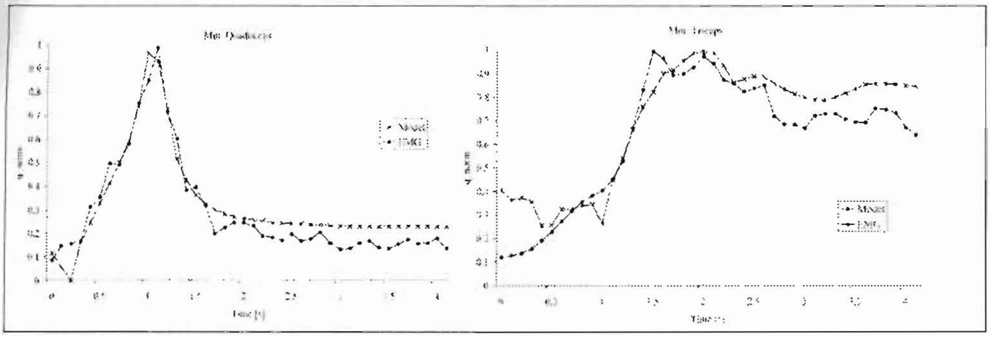


Figure 4: Comparison of measured and calculated muscle activation for L_70.

It is clear from Fig. 4 that the simulation results and measured data strongly correspond to each other. Similar correspondence was found for L_90 and SJ.

**CONCLUSION:** The advantage of this model is to gain information on muscle forces and muscle attributes which are not directly measurable. The verification of this a model was limited to comparing measured EMG data with the stimulation function. Good qualitative matches for the entire movement were found here. Another criterion was that the calculated muscle forces should never exceed either the measured isometric maximum force or the calculated F_poss. Therefore the values of the calculated forces seem to be appropriate as well as the force time histories regarding the stimulation function. The internal load of the Mm. Quadriceps was 6 to 10% higher than it was intended regarding the isometric maximum force gained in single joint tests. This could be due to the problem of comparing results of single joint test with multiple joint movements. In addition the model offers a possibility to improve the strength training by controlling the intensity of force development regarding the internal loads, which depend on muscle length and contraction velocity and not the "independent" external ones. Therefore further studies have do be done.

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