Human Multi-Body-System: Joint-resistance modeling based on muscle properties

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Introduction

In forensic biomechanics computer simulations may require a closer human-like dynamic behavior than elsewhere. The body-movements are not necessarily fast with respect to muscle action or reaction and the external forces may be well within physiological ranges. Thus, in multi-body–systems (MBS) applied to a diversity of forensic problems the modeling of joint properties gain essential weight.

The concern of this study is not the morphological correctness of joints, neither the modeling of the natural motion limitations, but the modeling of 'joint-resistance' in MBS. By 'joint-resistance', we understand not only the mechanical friction, but especially the elastic and non-elastic power which muscle-systems set against externally induced joint motion – however, without any active muscle contraction. In fact, it is the resistance, the muscles involved set against eccentric load (i.e. extension).

Methods

A 17-segment human MBS-Model was build using physical data provided by the Calcman 3D regression program (Henze, 1996). The program Simpack (Intec GmbH, D-82234 Wessling, Germany) was applied for the simulation calculations.

The ‘joint-resistance’ was modeled as the sum of two components:

– an elastic torque due to the passively extended muscles and
– a dissipative contribution by the mechanical friction.

The elastic torque due to the passively extended muscles is the product out of the cross-section based muscle-force, the lever arm, the muscle activity factor and an angle-dependent function describing the force-length-dependence of the extended muscle calculated for each muscle of the set.

The muscle force at isometric contraction as calculateed from the physilogical cross sectional area (PCSA) is:  

\[ F_0^M = \text{PCSA} \cdot 0.33 \cdot 10^6 \]  

(Veeger et al. 1995, Wülker et al. 1991)

Following a proposal of Zajac (1989), the muscle force, \( F^M \), was set to:  

\[ F^M = 1.5 \cdot F_0^M \]  

based on Zajac’s force-velocity-relation of the fully activated, eccentrically tensed muscle at optimum length. Introducing the effective lever arm, \( hba \), the relative activity \( q \) and the force-joint angle-relation, \( f(\alpha) \), the elastic component of the joint-resistance is described by:  

\[ M^M = F^M \cdot hba \cdot q \cdot f(\alpha) \]

Considering the passive force the muscles set against stretching load, only the right part (length > \( l_{u,0} \)) in the diagram of Zajac (1989), shown in Fig. 1, is valid. Based on this relation, the function \( f(\alpha) \) has been approximated by a straight line (Fig. 1) for convenient handling. As an example, the relative forces of the extensors and flexors of the knee joint set passively against lengthening is given in Fig. 2: The extensors expose the maximum resistance at an angle of 160°, the flexors at –5°. The intersection gives the neutral

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Fig. 1: Muscle-force and muscle-length (Zajac, 1989, modified), •••• linear approximation.

Fig. 2: Relative muscle-force and knee-angle.

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— Extensors, •••• Flexors.
position. This may be shifted either by stretching the abscissa to ‘virtual’ knee-angles or by setting the balance of the maximum forces of extensors and flexors accordingly. For the extensors ($e$) and flexors ($f$) of the knee-joint with a total range of 165° follows from Fig. 2

$$f_e(\alpha) = \left( \frac{1}{165} \alpha + \frac{160}{165} \right)$$

and

$$f_f(\alpha) = \left( -\frac{1}{165} \alpha + \frac{160}{165} \right)$$

resp.

The dissipative contribution includes the mechanical friction of the internal joint surfaces, the intraarticular fluid, and the damping of the surrounding tissue. According to Siff (1986), the dissipative moment is given by: $M_{\text{Diss}} = D \cdot \dot{\alpha}$, where $\dot{\alpha}$ stands for the angular velocity and $D$ for a constant in the range of 0.15 to 0.3, depending on the joint considered.

**Results and Discussion**

The model has been validated by the simulation of a 'knee-pendulum-experiment' of Riener (1997) The thigh was suspended on a plain table. The lower leg was released at an knee-angle of 66°. The MBS-model was build up using ellipsoids for thigh, lower leg and foot. The oscillations measured by Riener and the results of the simulation are given in Fig. 3.

![Fig. 3: Simulation of the 'knee-pendulum-experiment' of Riener (1997): bold = experimental oscillations thin = results of the simulation using $M_{\text{ext}} = 1.5 \cdot 5000 \cdot 0.05 \cdot 0.01 \cdot f_e(\alpha)$ [Nm] $M_{\text{flex}} = 1.5 \cdot 4000 \cdot 0.05 \cdot 0.01 \cdot f_f(\alpha)$ [Nm] $M_{\text{Diss}} = 0.3 \cdot \dot{\alpha}$ [Nm] (For further physical data of the model please contact the authors.)](image)

The ‘joint-resistance’ as described above has been introduced into the 17-segment human MBS-model. Two-dimensional full-size experimental falls on the knees have been performed and simulated to identify the parameters 'neutral position', 'force-length-dependence', and 'muscle activity'. Two different fall situations were encountered: Stiff knee-joints and ‘knee-fall’. Knee-pads and a gymnastic mattress were used for save landing of the volunteer. Knee-impact was registered by a load-plate. An example the recording of the motion analysis system (Motion Analysis Inc., Santa Rosa, Cal.) and the ‘visual’ results of the simulation of a ‘knee-fall’ are given in Fig. 4. In addition, the measured and calculated vertical head-velocity are recorded in Fig. 5 for comparison. Knee-touchdown is at 1.6 seconds. Correspondence is given until the volunteer prepares actively his landing at about 1.7 seconds. Further kinematic results validating the system will be published elsewhere.

**References**


Fig. 4: Man and model:
Above: Volunteer and recording of the motion analysis system, phases separated.
Below: Model simulation and superposition of model and experiment.

Fig 5: Experimental ‘Knee-fall’:
Measured (red) and calculated (black) vertical velocity of the head. Knee-touchdown at 1.6 seconds. Correspondence is given until the volunteer prepares actively his landing at about 1.7 seconds.

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