Influence of Ligament Modelling on Knee Joint Kinematics with respect to Multibody Optimisation

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Knee joint kinematics depends on complex interaction between the specific geometry of the articulating partners and adjacent soft tissues like ligaments and muscles. Ligament stiffness has great influence on knee stability and dynamics of the knee joint. In previous work [5] we identified tibiofemoral ligament parameters were identified by an optimisation procedure minimising the differences between measured and simulated tibiofemoral kinematics. Hereby a human cadaver experiment and a sophisticated musculoskeletal multibody model are used.

The objective of the present study is to investigate how the ligament bundles and the posterior capsule comprising up to 18 ligaments [5] can be described by a reduced set of four most important ligaments (MCL, LCL, ACL, PCL) [4] with respect to multibody optimisation. For this purpose, the influence of different ligament parameters on the knee joint kinematics are analysed by means of a musculoskeletal multibody model of the knee shown in Fig. 1a is used. The tibiofemoral joint is modelled by a polygonal contact model which enables simulation of the roll-glide movement [4]. Pelvis, femur, tibia/fibula and patella are modelled as rigid bodies. The geometric parameters were obtained from CT scans of an alcohol-fixed human leg specimen [2]. The patellofemoral joint is modelled by a user-defined one-degree-of-freedom joint that describes the path of the patella along a femur-fixed path with the arc length s, identified from a previous polygonal contact simulation of the patellofemoral joint. Wrapping of the quadriceps tendon was then implemented by a scleronomic constraint depending on the patella arc length s.

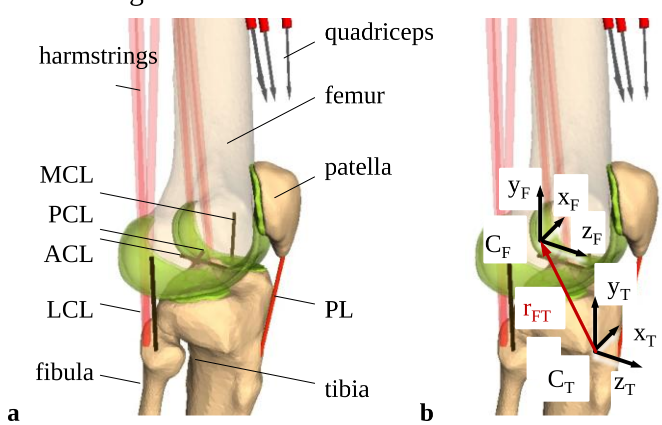

Fig. 1 **a** Multibody model of right knee (lateral view) with anterior and posterior cruciate lig. (ACL/PCL), lateral and medial collateral lig. (LCL/MCL), patellar lig. (PL). **b** Femur bone reference system C_F and tibia bone reference system C_T

The four most important ligaments, MCL, LCL, ACL and PCL according to [4], were modelled with non-linear dependencies of a ligament force f from the corresponding ligament strain ε according to [1]

$$f(\varepsilon) = \begin{cases} 0 & \varepsilon > 0 \\ 0.25k\dfrac{\varepsilon^2}{\varepsilon_0} & 0 \le \varepsilon \le 2\varepsilon_0 \\ k(\varepsilon - \varepsilon_0) & \varepsilon > 2\varepsilon_0 \end{cases} \quad \text{with} \quad \varepsilon = \frac{\ell - \ell_0}{\ell_0} \tag{1}$$

with the ligament stiffness k, the strain for the transition from quadratic to linear characteristic $\varepsilon_0 = 0.015$, the actual ligament length ℓ and the ligament zero load length ℓ_0.

The position of the femur bone system C_F relative to the tibia bone system C_T is described by the displacement vector $\boldsymbol{r}_\mathrm{FT} = [\Delta x \quad \Delta y \quad \Delta z]^\mathrm{T}$ and Cardan angles $\boldsymbol{\beta}_\mathrm{FT} = [\alpha \quad \beta \quad \gamma]^\mathrm{T}$ according to [3], see Fig. 1b. For N prescribed flexion angles α_i and given ligament parameters $\boldsymbol{p}$ the multibody model yields the five remaining displacement components

$$\boldsymbol{g}_i(\alpha_i, \boldsymbol{p}) = \left[\Delta x_i \quad \Delta y_i \quad \Delta z_i \quad \beta_i \quad \gamma_i\right]^\mathrm{T}, \; i = 1,\ldots,N. \tag{2}$$

The ligament parameters $\boldsymbol{p}$ are the stiffnesses and the reference strains of the four ligaments according to Fig. 1a, thus altogether eight parameters. The reference strain ε^ref is the strain in maximum knee extension. Assuming parameters from literature $\boldsymbol{p}^\mathrm{lit}$ [5] and modified parameters $\boldsymbol{p}^\mathrm{mod}$ with the corresponding displacements $\boldsymbol{g}_i^\mathrm{lit}(\alpha_i, \boldsymbol{p}^\mathrm{lit})$ and $\boldsymbol{g}_i^\mathrm{mod}\left(\alpha_i, \boldsymbol{p}^\mathrm{mod}\right)$, respectively, the sums of the squared differences of the displacement components are considered, examplary for the displacement component in medial-lateral direction Δx_i,

$$Z_x(\boldsymbol{p}^\mathrm{mod}) = \sum_{i=1}^{N} \left(\Delta x_i^\mathrm{mod} - \Delta x_i^\mathrm{lit}\right)^2. \tag{3}$$

As an example the influence of the stiffness of the MCL k_MCL on the displacement component Δx over the flexion angle α is shown in Fig. 2a. A significant dependence of Δx from k_MCL up to values of about 4000 N and almost no influence beyond is demonstrated. Then the medial-lateral movement is almost entirely restricted by the geometry of the articulating surfaces of the tibial and the femoral components. Figure 2b shows the influence of reference strain $\varepsilon_\mathrm{MCL}^\mathrm{ref}$ and stiffness k_MCL on the sums of the squared differences $Z_x(\boldsymbol{p})$ according to (3), whereby the remaining ligament parameters are kept fixed. The minimum zone around the literature values is stretched in direction of the stiffness.

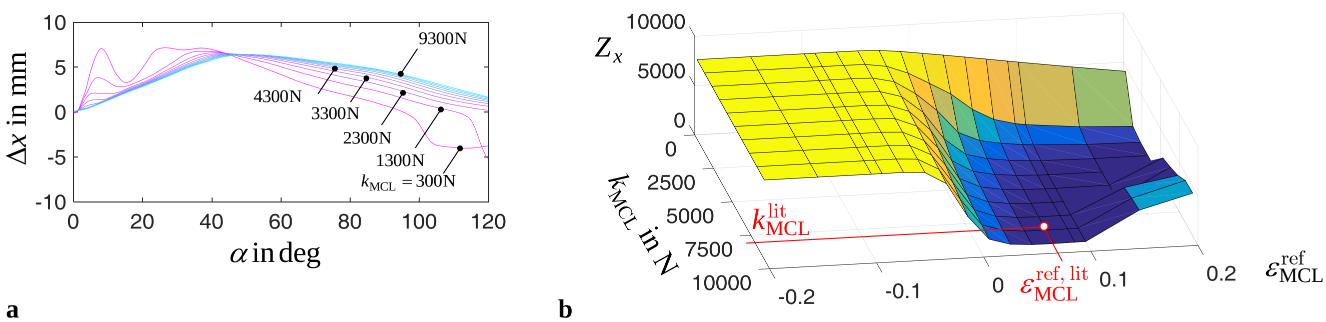

Fig. 2: Influence of the MCL on tibiofemoral kinematics. **a** Influence of stiffness k_MCL on the medial-lateral displacement Δx of C_F relative to C_T over the flexion angle α with the other ligament parameters kept constant. **b** Sums of the squared differences $Z_x(\boldsymbol{p}^\mathrm{mod})$ of the medial-lateral displacements Δx according to (3) under variation of stiffness k_MCL and reference strain $\varepsilon_\mathrm{MCL}^\mathrm{ref}$

In summary, the sums of the squared differences of all displacement components in (2) defined in analogy to (3) are used to build up the weighted objective function for identification of ligament parameters. The influences of the eight ligament parameters on the sums of the squared differences have been extensively investigated. The results help both to understand the parameter influences from a clinical point of view and to define appropriate weights of the objective function. We point out that the tibiofemoral kinematics of a complex system of ligaments can be properly described by the multibody model with the four knee ligaments ACL, PCL, MCL and LCL.

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