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Comparative assessment of knee joint models used in multi-body kinematics optimisation for soft tissue artefact compensation

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ABSTRACT

Estimating joint kinematics from skin-marker trajectories recorded using stereophotogrammetry is complicated by soft tissue artefact (STA), an inexorable source of error. One solution is to use a bone pose estimator based on multi-body kinematics optimisation (MKO) embedding joint constraints to compensate for STA. However, there is some debate over the effectiveness of this method. The present study aimed to quantitatively assess the degree of agreement between reference (i.e., artefact-free) knee joint kinematics and the same kinematics estimated using MKO embedding six different knee joint models. The following motor tasks were assessed: level walking, hopping, cutting, running, sit-to-stand, and step-up. Reference knee kinematics was taken from pin-marker or biplane fluoroscopic data acquired concurrently with skin-marker data, made available by the respective authors. For each motor task, Bland-Altman analysis revealed that the performance of MKO varied according to the joint model used, with a wide discrepancy in results across degrees of freedom (DoFs), models and motor tasks (with a bias between −10.2° and 13.2° and between −10.2 mm and 7.2 mm, and with a confidence interval up to ±14.8° and ±11.1 mm, for rotation and displacement, respectively). It can be concluded that, while MKO might occasionally improve kinematics estimation, as implemented to date it does not represent a reliable solution to the STA issue.

1. Introduction

Joint kinematics estimation commonly relies on methods involving a mechanical model of the locomotor apparatus together with the stereophotogrammetric tracking of skin-marker trajectories. However, soft tissue artefact (STA), i.e., the relative movement between the skin-markers and the underlying bones, introduces errors that jeopardise the information content of the skeletal motion estimation (Leardini et al., 2005; Peters et al., 2010). Since the artefact has a frequency content similar to that of bone movement, the problem cannot be solved by filtering (Chiari et al., 2005).

Multi-body kinematics optimisation (MKO) is increasingly used with the intent to compensate for STAs. The method embeds a rigid multi-body system and kinematic models of the joints involved, which means that the degrees of freedom (DoFs) of the joints are constrained (Andersen et al., 2009; Bonnechère et al., 2015; Charlton et al., 2004; Duprey et al., 2010; Lu and O'Connor, 1999; Ojeda et al., 2014; Reinbolt et al., 2005).

Various mechanical linkages representing the knee joint and embedded in MKO have been described. These involve major simplifications with respect to real and subject-specific joints, and have less than six independent DoFs. The hinge joint (Andersen et al., 2009; Reinbolt et al., 2005) allows rotation about only the flexion-extension axis. The spherical joint, the most common representation of the knee in MKO (Charlton et al., 2004; Lu and O'Connor, 1999; Ojeda et al., 2014), allows all rotational movements but no translation. These models provide, in most cases, a rather inadequate 3D representation of the physiological movement of the knee (Andersen et al., 2010; Clément et al., 2017). Parallel mechanisms have also been used (Duprey et al., 2010; Gasparutto et al., 2015; Valente et al., 2015), the principle of which relies on compound joints representing an assembly of simple mechanical linkages. Although these models generally allow most rotations and translations, they couple the DoFs thereby prescribing displacements in a deterministic way (i.e., kinematics is imposed by the geometry of the model). A different approach consists in representing the behaviour of the knee directly by

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mathematically coupling the DoFs (Bonnechère et al., 2015; Li et al., 2012; Scheys et al., 2011), with up to five DoFs driven by the flexion angle. A more recent modelling approach relies on a knee joint stiffness matrix and minimization of the relevant deformation energy (Richard et al., 2016).

Based on a number of studies assessing MKO, it may be concluded that no fully satisfactory knee joint model has been found yet (Andersen et al., 2010; Clément et al., 2017; Gasparutto et al., 2015; Richard et al., 2016). However, each of these assessment studies was performed on a single motor task (i.e., level walking, stepping-up, running or squatting). Moreover, some motor tasks (e.g., hopping, cutting) have not yet been investigated. Nor have all the above-mentioned joint models been compared to date. Finally, existing comparisons have been based on the root mean square error between estimated and reference kinematics, without performing any deeper analysis regarding the relevant degree of agreement (McLaughlin, 2013).

This study aimed to comprehensively compare the performance of MKO embedding six different knee joint models selected from those proposed in the literature. This was made possible thanks to the availability of concurrently acquired reference, virtually artefact-free, bone kinematics and skin-marker data (Cereatti et al., submitted for publication). The following motor tasks performed by able-bodied volunteers were analysed: level walking, hopping, cutting, running, sit-to-stand, and step-up. The degree of agreement between the reference and the MKO-estimated joint kinematics was assessed by Bland-Altman analysis as well as using the relevant root mean square error and determination coefficient.

2. Materials and methods

2.1. MKO framework

In this study, each bony segment is fully located and oriented (i.e., bone pose) in the global reference coordinate system by means of natural coordinates (de Jalon et al., 1994; Dumas and Chèze, 2007). Only the knee joint (i.e., the tibio-femoral joint) was considered in the study, meaning that the only segments involved in the MKO were the shank and the thigh.

Three types of constraints are typically used in MKO: driving constraints $\boldsymbol{\Phi}^m$, rigid body constraints $\boldsymbol{\Phi}^r$, and kinematic constraints $\boldsymbol{\Phi}^k$. The constraints are split into two sets of equations. A set of "soft" constraints contains the equations that may be violated (i.e., $\boldsymbol{\Phi}^m$). These constraints define the objective function of MKO, $f = (\boldsymbol{\Phi}^m)^T\boldsymbol{\Phi}^m$. A set of "hard" constraints contains the equations that must be fulfilled ($\boldsymbol{\Phi}^r$, $\boldsymbol{\Phi}^k$). In this framework, MKO is thus, to be regarded as a constrained optimisation problem. Note that a subset of the kinematic constraints $\boldsymbol{\Phi}^{k_1}$, especially in the case of ligaments, may be considered as "soft" constraints and appended in the objective function, $f = \begin{pmatrix} \boldsymbol{\Phi}^m \\ \boldsymbol{\Phi}^{k_1} \end{pmatrix}^T [\mathbf{W}] \begin{pmatrix} \boldsymbol{\Phi}^m \\ \boldsymbol{\Phi}^{k_1} \end{pmatrix}$ (with a weight matrix $\mathbf{W}$).

The present study considered the following knee joint models described in the literature and implemented, using natural coordinates, as kinematic constraints within MKO:

- *None:* no joint model, where the relative movement of the tibia and the femur are independent from each other and joint dislocation is therefore possible (this is, of course, a borderline case of MKO);
- *Spherical:* spherical joint model, allowing the three rotations while impeding the three displacements;
- *Hinge:* hinge joint model, allowing only one rotation about the flexion-extension axis while impeding the other DoFs;
- *Parallel:* parallel mechanism with minimized ligament length variation, where two sphere-on-plane contacts stand for the contact between femoral condyles and tibial plateau.
- *Coupling:* coupling curves between the DoFs, where internal rotation and adduction angles, as well as anterior and proximal displacements are functions of the extension angle through polynomial functions, and where lateral displacement is impeded;
- *Elastic:* elastic joint model based on the stiffness matrix, where all six DoFs are defined by the minimisation of the deformation energy.

A detailed description of the MKO method embedding the different models (i.e., kinematic constraints $\boldsymbol{\Phi}^k$) can be found in Duprey et al. (2010), Gasparutto et al. (2015) and Richard et al. (2016). More specifically, for model *Parallel*, the model geometry was taken from Parenti-Castelli and Sancisi (2013) and, for model *Coupling*, the coupling curves between the DoFs were an adaptation, due to a different sign convention, of those provided by Walker et al. (1988). Note that the MKO embedding model *None* is actually equivalent to a single-body optimisation (e.g., Soderkvist and Wedin, 1993).

2.2. Joint kinematics estimation

Joint coordinate systems used to compute the kinematics of the knee joint were defined so as to satisfy the conventions for axes and Euler sequence proposed by the ISB (Wu et al., 2002). The actual joint angles and displacements (extension, adduction, and internal rotation angles, and lateral, anterior, and proximal displacements) were computed from the natural coordinates (Dumas et al., 2012).

2.3. Experimental data

Right thigh and shank movement data from a single trial for each of the selected motor tasks, performed by able-bodied male subjects, were used for the analysis. These data were obtained from the datasets reported in Cereatti et al. (submitted for publication). They included both virtually artefact-free bone-pose data, obtained using either pin-markers or biplane fluoroscopy, and concurrently acquired skin-marker data. The bony segment coordinate systems were defined based on bone anatomy and the reference positions of the skin-markers with respect to these coordinate systems were defined as their mean positions over the duration of the motor task. The data for level walking, hopping and cutting were from one volunteer (age: 22 years, mass: 63 kg, height: 1.75 m; Benoit et al., 2006), while the data for running were from another volunteer whose anthropometric features were unknown (Reinschmidt et al., 1997). Relevant artefact-free data were obtained using pins inserted in the distal femur and proximal tibia. Data for the step-up and sit-to-stand tasks were from one male volunteer (age: unknown, mass: 83 kg, height: 1.75 m) and artefact-free data were obtained via biplane fluoroscopy (Tsai et al., 2011). Further details concerning the experimental set-ups, the definition of bony segment coordinate systems and relevant calibration and registration procedures used for the different datasets can be found in Cereatti et al. (submitted for publication) and in the above-mentioned references.

2.4. Assessment

For each of the six motor tasks, reference (i.e., artefact-free) femur and tibia pose and knee joint angles and displacements were reconstructed using pin-marker or biplane fluoroscopy data. Femur and tibia pose and knee joint kinematics were also estimated using the concurrently acquired skin-marker data and six MKO procedures each embedding one of the above-illustrated knee joint models (*None*, *Spherical*, *Hinge*, *Parallel*, *Coupling* and *Elastic*). The degree of agreement between the joint angles and displacements derived from the six MKO procedures and the reference kinematics was assessed through Bland-Altman analysis (Bland and Altman, 1986). The bias (b) and confidence interval (CI; i.e., 1.96 standard deviation) were calculated. The root mean square error ($RMSE$) and coefficient of determination (R^2) were also calculated for the sake of comparison with previous studies. Note that, when using the models *Spherical* and *Hinge*, displacements were null, thus the relevant coefficient of determination could not be computed.

3. Results

Full results are presented here for three motor tasks: level walking, hopping and cutting, while results for the other motor tasks are reported in Supplementary Material. Note that the results reported in the body of the paper and those in Supplementary Material lead to the same general conclusions.

3.1. Kinematics

Both reference joint angles and displacements and those estimated using MKO embedding the six joint models, are represented in Figs. 1–3 for level walking, hopping, and cutting, respectively. Overall, the extension angles estimated through the six MKO procedures and the reference angles exhibited similar patterns. However, conflicting results emerged with regard to the other DoFs. Depending on the motor task and the DoF considered, large discrepancies were observed in the kinematic outcomes of the MKO for any given joint model. Note that the characteristics of the knee models meant that model *Spherical* provided null displacements, model *Hinge* provided null displacements as well as null adduction and internal rotation angles, and model *Coupling* provided null