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Knee joint kinematics: comparison of two optimization models with respect to data noise

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The interpretation of human knee joint kinematics in terms of displacements is an outcome of the underlying model of the joint and the measurement technique. Measurement errors and noise challenge the development of optimization procedures which, based on a reduction in degrees of freedom, aim for the reproducibility of joint displacements by computational techniques. So far, optimization algorithms have been applied which are based on a kinematic model of the healthy human tibio-femoral joint (TFJ) as a compound hinge with two fixed orthogonal axes. On the other hand, empirical studies find non-orthogonal rotational axes. Therefore, it was the aim of the present study to investigate the implications of a refined kinematic model on the accuracy of computed joint rotation angles. For the purpose of quantitative comparison, kinematic data of a TFJ with two axes intersecting at an arbitrary angle were simulated. The joint rotations were optimized for the assumption of (a) two orthogonal and intersecting axes (model A), and (b) two axes intersecting at an arbitrary angle (model B). Model B recovers the original input data closer in case of low noise level as encountered in invasive measurement techniques. Skin mounted markers tracking involves non-normally distributed noise which is typically larger by on order of magnitude. In this case, model A exhibits a more favorable performance. These observations motivate the search for alternative kinematic descriptions of the TFJ.

1 Tibio-femoral joint kinematics

The relevance of understanding knee joint kinematics reveals itself in a number of clinical topics. For instance, prostheses have to be designed such that the flexibility and stability characteristics of the human knee joint are reproduced as close as possible in order to limit loosening and accelerated wear. Anatomically, the human knee joint is composed of two joints, the patellofemoral and the tibiofemoral joint (TFJ), the latter to which the present study is devoted to. In three dimensions, a linkage between two rigid bodies like femur and tibia exhibits six degrees of freedom (DOF), three rotational and three translational DOFs. But [1] argue that, in many clinical applications, accounting for full 6 DOFs knee joint motion data is not required. Moreover, it is generally difficult to associate the kinematic data obtained from movement analysis with displacements with respect to the clinical directions. Especially, non-invasive measurement techniques with skin-mounted markers involve large measurement errors due to soft tissue artifacts, e.g., [2, 3]. Thus, in a computational approach one aims for compensating errors by fitting kinematic data to a simplified model of the TFJ with the purpose of a reliable identification of joint angles,.

The TFJ has been intensively studied in the past, involving a paradigmatic shift in the understanding of human knee joint kinematics. Based on observations of the flexion cycle in the sagittal plane, i.e., in two dimensions, it was concluded that the flexion-extension (FE) axis moves within the posterior femoral condyle. However, [4, 5] found that the plane of the FE movement and the sagittal observation plane were tilted by about 5-10° implying that even a fixed axis would appear to move as an artefact of the procedure ([6, 7] and Refs. therein). The hypothesis of an, in the posterior femoral condyles fixed, FE axis was supported by three-dimensional cadaveric studies revealing the existence of isometric points on the distal femur [6, 7, 8]. A slice of the femur in a plane orthogonal to the line connecting the isometric points, exhibits circular shaped femoral condyles [9]. Further in vitro research identifies two FE-axes, one acting in the regime of 150° to 15° of knee flexion and one in the regime until full extension ([8], [6] and Refs. therein). TFJ motion may be represented as a rotation around two independent axes, first, the femur fixed FE axis and, second, a tibia-fixed (longitudinal) rotation axis (TR axis) [7]. In this model, femoral roll back is the manifestation of TR rotation, also called internal-external (ie) rotation, during knee flexion. The location of the TR axis varies as the movement is not only governed by the bony structure but also by muscles, ligaments, and load bearing forces [6]. The FE axis seems to be fixed; though, there exist several different definitions of this axis on the basis of anatomical landmarks [7, 9, 10]. Furthermore, the acquisition of anatomical landmarks by palpation introduces a quite significant error and high intersession variabilities. Thus, it is aimed to replace landmark based axes by functional rotational axes calculated from kinematic data.

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2 Computation of functional joint axes

During load bearing flexion, it was observed that the flexion interval of 40°–80° is dominated by pure flexion. Between 0° and 40°, flexion is accompanied by tibial rotation. Therefore, [11, 12, 13, 14, 15] fix the FE axis in the former interval. Assuming the FE and the TR axis to be perpendicular and intersecting, the TR axis is extracted by minimizing the varus-valgus rotation within the flexion cycle (model A). As empirical studies find non-orthogonal axes [16], the present study introduces an analogous procedure simultaneously optimizing the angle between the two axes (model B). The precise details are described in [17], where – for the purpose of a quantitative comparison – models A and B are applied to simulated kinematic data of a TFJ with an arbitrary angle between the two axes. The input parameters of the kinematic data simulation comprise the (time invariant) 3D-angle (α, γ) between the FE and the TR axis and the time dependent angles $\theta(t)$ (FE rotation) and $\phi(t)$ (TR or ie rotation). Experimental data was used as input for $\phi(\theta)$ [18, 19]. Different mutual orientations of the two axes were implemented by varying α. Finally, normally distributed noise of the size of typical measurement errors was included.

Our optimization procedure was designed to aim for the computational reconstruction of the input parameters – $\gamma, \theta(t), \phi(t)$ (model A: $\alpha = 90°$), or $\alpha, \gamma, \theta(t), \phi(t)$ (model B). The number of unknowns were reduced by parameterizing $\theta(t)$ and $\phi(t)$. As expected, in the absence of noise model B recovered the input parameters exactly, model A only for $\alpha_{\text{input}} = 90°$. With increasing noise, model A performed better compared to model B.

3 Discussion

The two models confronted describe a joint with intact ligaments and in the absence of osteoarthritis. An injured knee may exhibit further rotational and translational DOFs not accounted for in either of the models. An adequate model should match the real kinematics as close as possible; however, a compromise has to be found – balancing two concurring aims, (a) sophistication of the kinematic model and (b) robustness with respect to data noise.

In our simulations, the objective function of model A exhibits a clearly pronounced minimum. For model B, a ‘flat valley’ in some of the optimized parameters is faced near the minimum: Noise distorting the shape of the objective function causes the optimization process to be mildly ill-posed, i.e., the 4-parameter identification problem tends to be underdetermined. The conclusion of this study is that, small noise as present in medical imaging techniques or bone pins in cadaver measurements can be well treated using model B permitting a closer insight into TFJ kinematics. Larger errors as resulting from soft tissue artifacts demand the stricter orthogonality constraint of model A. Future investigations will be devoted to the improvement of the modeling process, and also testing with experimental kinematic data of human knee joints.

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