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Evaluation of knee functional calibration with and without the effect of soft tissue artefact

Morgan Sangeux [a,b,c,*], Arnaud Barré [d], Kamiar Aminian [e]

[a] The Royal Children's Hospital, Melbourne, Australia
[b] The Murdoch Childrens Research Institute, Australia
[c] The University of Melbourne, School of Engineering, Australia
[d] Laboratoire de recherche en Imagerie et Orthopédie (LIO), ETS, Montreal, Canada
[e] Ecole Polytechnique Fédérale de Lausanne, Lausanne, Switzerland

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ABSTRACT

Functional calibration methods were devised to improve repeatability and accuracy of the knee flexion–extension axis, which is used to define the medio-lateral axis of the femur coordinate system in gait analysis. Repeatability of functional calibration methods has been studied extensively in healthy individuals, but not accuracy in the absence of a benchmark knee axis. We captured bi-plane fluoroscopy data of the knee joint in 17 subjects with unilateral total knee arthroplasty during treadmill walking. The prosthesis provided a benchmark knee axis to evaluate the functional calibration methods. Stereo-photogrammetry data of thigh and shank marker clusters were captured simultaneously to investigate the effect of soft tissue artefact (STA). Three methods were tested, the Axis Transformation Technique (*ATT*) finds the best single fixed axis of rotation, *2DofKnee* finds the axis that minimises knee varus–valgus and *trajAJC* finds the axis perpendicular to the trajectory, in the transverse plane of the femur, of a point located on the longitudinal axis of the tibia. Using fluoroscopy data, functional axes formed an angle of less than 2° in the transverse plane with the benchmark axis. True internal–external range of movement was correlated with decreased accuracy for *ATT*, while varus–valgus range of movement was correlated with decreased accuracy for *2DofKnee* and *trajAJC*. STA had negative impact on accuracy and variability. Using stereo-photogrammetry data, the accuracy of *2DofKnee* was 1.7°(SD: 5.1°), smaller than *ATT* 2.9°(SD: 5.1°) but not to *trajAJC* 1.7°(SD: 5.2°). Our results confirm that of previous studies, which utilised the femur condylar axis as reference.

1. Introduction

Identifying the correct orientation of the medio-lateral axis of the femur is key to the accuracy and utility of gait analysis. The medio-lateral axis of the femur is the secondary axis used to define the anatomical femur coordinate system, and is orthogonal to the primary axis determined by the ipsilateral hip and knee joint centres. The orientation of the medio-lateral axis determines hip rotation kinematics during gait, which is a key variable to inform derotation osteotomies in children with cerebral palsy (Schwartz et al., 2014). The orientation of the medio-lateral axis also affects knee kinematics during gait (Schache et al., 2006) which is a key outcome measure in total knee arthroplasty (Levinger et al., 2012; McClelland et al., 2007).

However, the medio-lateral axis of the femur is difficult to determine accurately and reliably. The conventional gait model requires assessors to palpate anatomical landmarks, the medial and lateral epicondyles of the femur, or to use a knee alignment device which clamps on these bony landmarks (Baker, 2013). The localisation of these bony landmarks may be difficult and only accurate to within centimetres (Della Croce et al., 2005). Errors in marker placement may lead to large differences in the orientation of the femur medio-lateral axis because the epicondyles are only a short width apart, typically 10 cm.

Functional calibration offers an alternative to determine the medio-lateral axis for gait analysis (Passmore and Sangeux, 2016). Contrary to conventional models, functional calibration does not rely on bony landmarks but uses the movement of the tibia with respect to the femur to determine the knee axis, which is used as a proxy for the medio-lateral axis of the femur either explicitly *e.g.*, Schache et al. (2006) or implicitly *e.g.*, Schwartz and Rozumalski (2005). Most functional calibration algorithms model the knee as a hinge joint, *i.e.*, with one degree-of-freedom (DoF), and an

* Correspondence to: The Hugh Williamson Gait Analysis Laboratory, The Royal Children's Hospital, 50 Flemington Road, Parkville, 3152 VIC, Australia.
E-mail address: morgan.sangeux@gmail.com (M. Sangeux).

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optimisation function is used to find a single axis of rotation (Ehrig et al., 2007). Some algorithms allow the knee to have 2 DoF, flexion–extension and internal–external rotation, for example *DynaKAD* (Baker et al., 1999; Rivest, 2005). To the best of our knowledge, no study has investigated the effect the modelling constraints (1 or 2 DoF) have on the accuracy of the resulting medio-lateral axis of the femur.

Soft tissue artefact (STA) affects the accuracy in tracking the movement of body segments, including the thigh and shank segments (Peters et al., 2010). Consequently, all functional calibration algorithms are affected by STA. The effect of STA on functional calibration algorithms has been estimated *in silico*, where Gaussian noise was added to the marker trajectories (Ehrig et al., 2007). Excellent accuracy ($<1°$) was found providing a range of movement (ROM) at the knee exceeding 45°. However, Gaussian noise may not be representative of STA (Dumas et al., 2015). Currently, there is limited data on the effect of STA on knee functional calibration *in vivo*.

Recent studies have provided results on functional calibration accuracy in healthy subjects with respect to an anatomical medio-lateral axis defined from freehand 3D ultrasound (Passmore and Sangeux, 2016) or from bi-plane x-ray imaging (Sauret et al., 2016). However, the results of these studies could not differentiate between the impact of the underlying knee model and the impact of STA at the thigh and shank segments. The aims of this study were therefore (1) to evaluate various functional calibration methods against a gold standard knee axis without the effect of STA and (2) to investigate the effect of STA at the thigh and shank segments on knee angles and on the performance of functional calibration methods.

2. Materials and methods

We performed a secondary analysis from data collected on 19 subjects equipped with unilateral total knee prosthesis (F.I.R.S.T, Symbios, Switzerland) (Barre et al., 2013, 2015). The prosthesis featured highly congruent articulation surfaces between the femoral component and the tibial polyethylene insert leading to a single flexion–extension axis between 0° and 90° flexion. The prosthesis also allowed knee internal–external rotation thanks to a single vertical axis flat-on-flat articulation between the tibial polyethylene insert and the tibial component. Subjects were equipped with a large number of markers on the thigh and the shank segments. The movement of the femoral and tibial components of the knee prosthesis were captured with bi-plane fluoroscopy and the marker clusters on the thigh and shank were captured simultaneously with stereo-photogrammetry. Two fluoroscopes were positioned with an angular separation of 50° so that a clear view of the knee joint was obtained while the subjects walked on a treadmill (Barre et al., 2013). Fluoroscopy data were captured at 30 Hz and synchronised with a 7 camera (MX3+) VICON system (Oxford metrics, UK). The motion capture system acquired the 3D position of the thigh and shank marker clusters at 240 Hz. The trajectory of the prosthesis from fluoroscopy was interpolated from 30 Hz to 240 Hz.

Clinical gait analysis entails the use of a reduced number of markers compared to the clusters used in the original study (Barre et al., 2013). We chose 5 markers from the thigh and shank segments (Fig. 1). For the thigh segment, the markers were chosen as follows: two located mid-thigh on the iliotibial band (THLP, THLD), 2 located mid-thigh on the anterior aspect of the thigh (THAP, THAD) and one marker approximately located over the lateral epicondyle of the femur (KNE). This choice is coherent with previous marker set used in clinical gait analysis (Sangeux et al., 2011). For the shank segment, we chose markers located approximately over the lateral and medial malleoli (ANK, MED), one marker located mid-shank and lateral (TIB), and two markers located on the tibial crest (TIAP, TIAD). This choice replicate the marker set described in a previous clinical gait analysis study (Peters et al., 2009). These thigh and shank marker clusters match the current gait analysis protocol at The Royal Children's Hospital, Melbourne, Australia.

The anatomical coordinate system for the femur and the tibia corresponded to the respective component of the prosthesis (Fig. 1). For the femur, the *X*-axis was medio-lateral, pointing to the left and determined from the articular surface of the prosthesis. According to the prosthesis design, this axis prescribed the flexion–extension axis between 0° and 90° of knee flexion. The *Y*-axis was the longitudinal axis pointing up and the *Z*-axis the anterior–posterior axis. For the tibia, the *X*-axis was the medio-lateral axis, pointing to the left and aligned with the tibial plateaux. The *Y*-axis was the longitudinal axis pointing up and the *Z*-axis

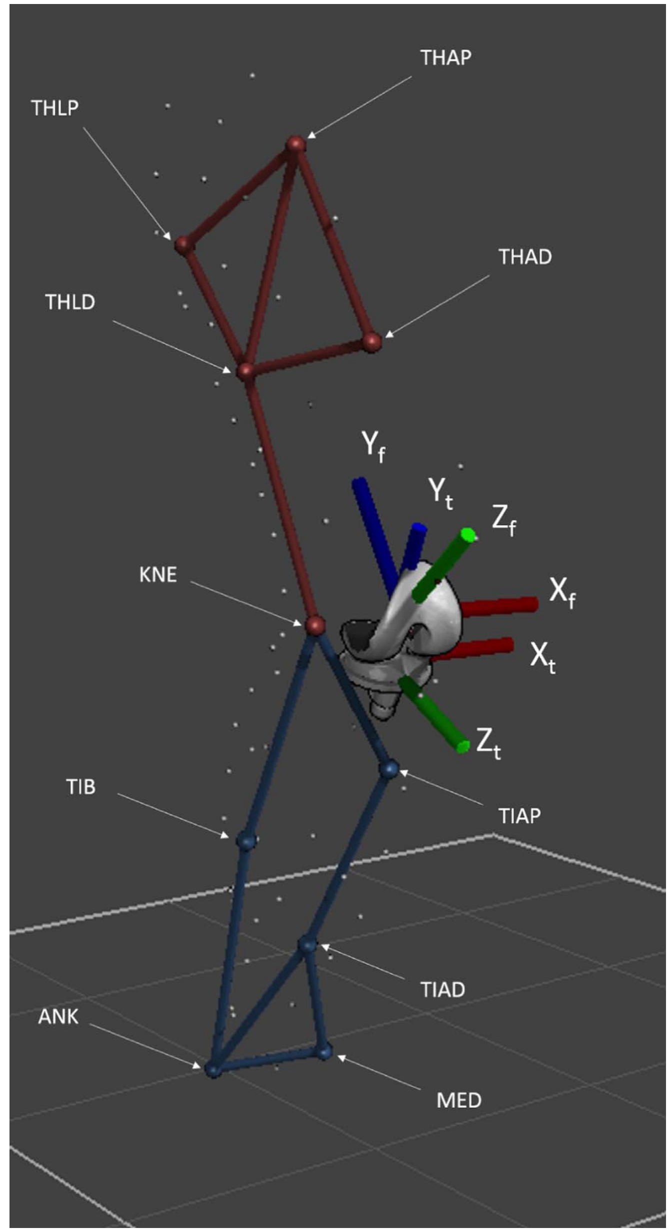

Fig. 1. Marker clusters and knee prosthesis for the right limb of subject #4. Clinical marker clusters are highlighted in red (thigh) and blue (shank) while all other markers are shown in white. Coordinate systems for the femoral (X_f, Y_f, Z_f) and tibial (X_t, Y_t, Z_t) components of the knee prosthesis are superimposed. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

was the anterior–posterior axis. The position and orientation of the coordinate system for the femur and tibia components were obtained from bi-plane fluoroscopy (Barre et al., 2013, 2015).

Static calibration was performed with the subjects standing still. The position of the thigh and shank marker clusters with respect to the femoral and tibial components of the prosthesis were recorded in the standing posture. During walking, we determined the rigid transformation between the static position and the new position of the marker clusters for each time frame using a least squares fitting approach (Söderkvist and Wedin, 1993). The rigid transformation of the thigh and shank marker clusters was applied to the static position of the femur and tibia components of the prosthesis to calculate the knee movement as obtained from stereo-photogrammetry, *i.e.*, contaminated by STA.

We used data from three consecutive strides in the centre of the walking trial (Passmore and Sangeux, 2016). The Cardan angles for the movement of the femur, tibia and knee (*i.e.*, tibia with respect to femur) were obtained from the floating axis sequence *X*- (flexion–extension), *Z*- (varus–valgus) and *Y*- (internal–external rotation) (Grood and Suntay, 1983). The movement of the femur

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