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Technical note

# Estimated landmark calibration of biomechanical models for inverse kinematics

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ABSTRACT

Inverse kinematics is emerging as the optimal method in movement analysis to fit a multi-segment biomechanical model to experimental marker positions. A key part of this process is calibrating the model to the dimensions of the individual being analysed which requires scaling of the model, pose estimation and localisation of tracking markers within the relevant segment coordinate systems. The aim of this study is to propose a generic technique for this process and test a specific application to the OpenSim model *Gait2392*. Kinematic data from 10 healthy adult participants were captured in static position and normal walking. Results showed good average static and dynamic fitting errors between virtual and experimental markers of 0.8 cm and 0.9 cm, respectively. Highest fitting errors were found on the epicondyle (static), feet (static, dynamic) and on the thigh (dynamic). These result from inconsistencies between the model geometry and degrees of freedom and the anatomy and movement pattern of the individual participants. A particular limitation is in estimating anatomical landmarks from the bone meshes supplied with *Gait2392* which do not conform with the bone morphology of the participants studied. Soft tissue artefact will also affect fitting the model to walking trials.



## 1. Introduction

Inverse kinematics [1–3], also known as global optimisation [4] and kinematic fitting, is emerging as the optimal method for fitting a multi-segment biomechanical model to a set of experimental marker positions measured during movement analysis. The quality of the kinematic fit is dependent on how a generic model of the skeletal system and the locations of markers upon it are adapted to reflect the anthropometry of the individual being measured and the actual positions of those markers. This will be referred to as model calibration. It is generally based on measurements of marker trajectories captured during a static trial with the person standing in a standardised pose. Several recent papers [5–8] have shown how susceptible model outputs can be to model scaling. Application to data from the Grand Challenge Competition to Predict In Vivo Knee Loads [9] suggests that *anatomical* calibration lead to better prediction of knee loads than *linear* scaling (and that *kinematic scaling* performs even better). OpenSim [10] is becoming more and more widely used for musculoskeletal modelling but, although it includes a *Scaling Tool*, this requires subjective interaction from the operator for each subject to complete full model calibration [11] by adjusting the virtual markers and marker weighting for a better fit. A subjective and variable process is thus introduced into a processing pathway which should be objective and standardised. This technical note will present an implementation of a generic model calibration technique within OpenSim using the Scaling Tool in such a way that subjective operator interaction is not required.

## 2. Methods

### 2.1. General principles

Model calibration involves three interacting processes; scaling (adjusting the dimensions of the model segments to correspond to those of the person being analysed), fitting (adjusting the pose of the model to that in which the person is standing) and model marker localisation (specifying the position of markers within the model to correspond to the measured position of actual markers). It is because these processes interact that the current OpenSim scaling tool needs to be applied iteratively. The key to the alternative proposed in this paper is to structure the operation in such a way that the interactions are removed such that the three processes can be applied in a single sequence which requires no user intervention.

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**Table 1.**
Segment scaling factors in its principal axis as well as axes in both other directions for all ten participants.

| | | S01 | S02 | S03 | S04 | S05 | S06 | S07 | S08 | S09 | S10 | Mean (SD) |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| **Pelvis** | A–P / P–D | 1.11 | 1.35 | 1.03 | 1.11 | 1.24 | 1.00 | 1.15 | 0.96 | 1.06 | 1.11 | 1.11 (0.12) |
| | M–L | 1.10 | 1.05 | 1.02 | 1.06 | 1.03 | 0.98 | 1.10 | 1.01 | 1.14 | 0.99 | 1.05 (0.05) |
| **Femur** | A–P / M–L | 1.37 | 1.66 | 1.33 | 1.32 | 1.65 | 1.17 | 1.82 | 1.40 | 1.22 | 1.23 | 1.42 (0.22) |
| | P–D | 0.91 | 1.00 | 1.07 | 1.12 | 1.01 | 1.03 | 0.91 | 1.04 | 1.16 | 1.10 | 1.04 (0.08) |
| **Tibia** | A–P / M–L | 0.75 | 1.02 | 0.85 | 0.81 | 0.80 | 0.91 | 0.76 | 0.77 | 0.83 | 0.82 | 0.83 (0.08) |
| | P–D | 0.91 | 1.03 | 0.98 | 1.00 | 0.99 | 0.94 | 0.95 | 0.99 | 1.04 | 0.98 | 0.98 (0.04) |
| **Foot** | A–P | 0.87 | 1.06 | 1.02 | 0.99 | 1.00 | 0.99 | 0.93 | 0.95 | 1.01 | 1.01 | 0.98 (0.05) |
| | P–D / M–L | 0.80 | 1.19 | 1.11 | 0.99 | 0.90 | 1.05 | 0.93 | 0.98 | 1.03 | 1.08 | 1.01 (0.11) |
| **Torso** | A–P / M–L | 0.95 | 1.13 | 1.09 | 1.10 | 1.00 | 1.16 | 1.00 | 0.96 | 1.19 | 1.11 | 1.07 (0.09) |
| | P–D | 0.91 | 0.95 | 0.87 | 1.01 | 0.88 | 0.85 | 0.83 | 0.93 | 0.97 | 0.84 | 0.90 (0.06) |

A–P = anterior–posterior, P–D = proximal–dorsal, M–L = medial–lateral.

The process of scaling is based on the position of a number of bony landmarks and joint centres (collectively referred to as *anatomical landmarks* below) which are both specified within the model (Table A1, on-line supplementary information) and whose position is estimated from measured marker trajectories (Table A2, on-line supplementary information). The process is essentially that of fitting the landmarks in the model to those estimated positions. This process, based on the standard scaling tool of OpenSim, has four stages operating on data captured during a static calibration trial*: landmark position estimation*, *model scaling*, *model fitting,* and *tracking marker localisation*.

1 *Landmark position estimation*. The positions of the anatomical landmarks are estimated based on the positions of skin mounted calibration markers in such a way that scaling and fitting can be performed simultaneously in a manner that is consistent with the biomechanical model (Table A1 and Fig. A1, on-line supplementary information). Anatomical landmarks are estimated on the basis of the measured calibration marker position, the known dimensions of the marker, an estimate of the soft tissue thickness (STT) and the known geometry of the model. Thus, for example, the lateral femoral epicondyle landmark is estimated to be offset by the marker radius, the thickness of the base plate and the estimated STT from the measured centre of the marker in the direction of the measured centre of the medial epicondyle marker. Joint centres are then estimated in relation to these landmarks (e.g. the knee joint centre is the mid-point of the medial and lateral epicondyle markers).
2 *Model scaling*. Each segment is then scaled separately along its principal axis [12] (longitudinal axis for foot, tibia, femur, thorax and medio-lateral axis for pelvis) and in the plane perpendicular to that (equally in both directions). Scaling along the principal axes fits the model segment to the distance between the estimated joint centres (e.g. hip and knee joint centres for the femur). Scaling in the plane perpendicular to this fits it to the estimated bony landmark (e.g. lateral and medial epicondyles for the femur).
3 *Model fitting*. Once the segments have been scaled the whole model is fitted to the marker data to minimise a weighted root mean square of the distances between modelled and measured anatomical landmarks. The weighted sum is used to bias the fit in favour of the joint centres.
4 *Tracking marker localisation*. Finally, the locations of the tracking markers referred to the relevant segment coordinate systems are calculated and recorded within the model for use when fitting the model to data from walking (or other movement) trials.

### 2.2. Specific implementation

A study to implement and test this technique was granted ethical approval by the College of Health and Social Care Ethics Panel. A convenience sample of ten healthy adult volunteers with no history of neuromusculoskeletal impairments was recruited from the university community ($28 \pm 5$ years old, $1.72 \pm 0.08$ m, $69 \pm 12$ kg) and gave informed consent before the measurement. The 34 markers listed in Table A2 (on-line supplementary information) were placed, which were adapted from the provided marker set in the OpenSim example data of *Gait2392*. A ten camera motion capture system (Nexus 1.8.5, Vicon, T40S cameras) was used to capture data from a static calibration trial and then from several walks at free walking speed over a walkway with four force plates (Kistler, $2 \times 9286$A, $2 \times 9253$A).

Marker locations were reconstructed within Nexus, while estimated anatomical landmarks and joint centres were calculated using a BodyLanguage model (Vicon, Oxford, UK) running within Nexus (Table A1, on-line supplementary information). Detailed description of additional landmarks and joint centres can be found in the on-line supplementary information.

This study was part of a project to validate a pipeline using reduced residual analysis [RRA, 13]. The generic musculoskeletal model to be scaled was the *Gait2392* model provided by OpenSim [14]. The model was mainly used with its standard settings, while estimated anatomical landmarks were, besides the experimental marker, additionally placed on the bony meshes of the model (Table A3, on-line supplementary information). The model was further slightly altered by locking the sub-talar and metatarsophalangeal (MTP) joints in a neutral position [15].

Scaling, fitting and tracking marker localisation were implemented using the OpenSim *Scaling Tool.* Compared to the proposed scaling tool of the example gait data of OpenSim, only the anatomical landmarks (not the original marker positions) were used for scaling and fitting to enable a better scaling along the axes of the segments. All participants were slim and lean adults, therefore, a small generic value for STT of 0.5 cm has been chosen. Segments were scaled separately along the principal axis to fit to joint centres and in the planes perpendicular to this to fit landmark pairs as specified in Table A4 (on-line supplementary information). The only exception to this was at the foot which was scaled vertically to fit the distance between the ankle joint centre and the floor and in the sagittal plane to fit the distance between the heel and toe markers. A weighting of 3:1 was in a preliminary study shown to be suitable for fitting to the principal axes to the joint centres leaving the landmark information to determine the rotation of segments about those axes.

Once the model had been scaled the *Inverse Kinematics* tool was used to fit it to one of the walking trials for each volunteer by minimising the RMS distance between modelled and measured tracking markers (Table A1, on-line supplementary information) with equal weighting on all markers. (Note that only markers and neither model nor measured landmarks are used in tracking the dynamic trials.) The mean and maximum RMS fitting error calculated across the trial was recorded for both the static and walking trials

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