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# Bone position estimation from skin marker co-ordinates using global optimisation with joint constraints

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**Abstract**

Widespread use of gait or motion analysis in the diagnosis of patients with locomotor pathology and the subsequent planning and assessment of treatment has been limited because of its reliability, particularly in evaluating frontal and transverse plane components. This is because spatial reconstruction of the musculoskeletal system and calculation of its kinematics and kinetics via a skin marker-based multi-link model are subject to marker skin movement artefacts. Traditional methods treat each body segment separately without imposing joint constraints, resulting in apparent dislocations at joints predominantly because of skin movement artefacts. An optimisation method for the determination of the positions and orientations of multi-link musculoskeletal models from marker co-ordinates is presented. It is based on the minimisation of the weighted sum of squared distances between measured and model-determined marker positions. The model imposes joint constraints. Numerical experiments were performed to show that the new method is capable of eliminating joint dislocations and giving more accurate model position and orientation estimations. It is suggested that, with joint constraints and a global error compensation scheme, the effects of measurement errors on the reconstruction of the musculoskeletal system and subsequent mechanical analyses can be reduced globally. The proposed method minimises errors in axial rotation and ab/adduction at the joints and may extend the applicability of gait analysis to clinical problems. © 1999 Elsevier Science Ltd. All rights reserved.

*Keywords:* Bone position; Multi-link models; Kinematics; Optimisation; Joint constraints

## 1. Introduction

Widespread use of gait or motion analysis in the diagnosis of patients with locomotor pathology and the subsequent planning and assessment of treatment has been limited because of its reliability, particularly in evaluating frontal and transverse plane components. This is critical because, in patients with pathological gait, such as children with cerebral palsy, abnormalities occur essentially in these planes (Gage, 1991). In skin marker-based gait analysis systems, skin movement artefacts have been shown to affect the accuracy of calculated joint kinematics much more in the frontal and transverse planes than in the sagittal plane (Cappozzo et al., 1996). Therefore, reduction of the effects of skin movement artefacts in the two planes will improve the quality of gait analysis data for clinical purposes.

The use of video-based stereophotogrammetry in human movement analysis requires determination of the poses (position and orientation) of the body segments from skin-mounted markers before their kinematics and kinetics can be calculated. The musculoskeletal system is generally modelled as a multi-link chain with each body segment as a rigid link. An array of at least three markers per segment is needed for the definition of a segment-embedded reference frame which represents the pose of the segment. Due to skin movement, the marker array displaces and rotates rigidly relative to the underlying bone. Furthermore, the shape of the array changes. Both effects introduce errors into the estimated segment pose. Efforts have been made to improve measurement techniques to minimise skin movement artefacts (Cappello

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et al., 1997) but they cannot be eliminated unless markers are applied to the bones directly or through bone-pins (Fuller et al., 1996; Lafortune et al., 1992). Therefore, spatial reconstruction of the musculoskeletal system and calculation of its kinematics and kinetics via a skin marker based multi-link model should take account of skin movement artefacts.

Poses of multi-link models for gait studies are often obtained by calculating sequentially the separate pose of each segment, without considering joint constraints. Several methods for estimating the pose of a single segment have been proposed. One frequently used method, here referred to as the *direct method* (*DM*), calculates the segment-embedded frame from two vectors, pointing from one of the three markers to the other two (e.g. Apkarian et al., 1989; Kadaba et al., 1990). Skin movement artefacts are not considered. Another type of method, here referred to as the *segmental optimisation method* (*SOM*), estimates the segment pose in terms of its transformation matrix by minimising marker array deformation from its reference shape in a least-squares sense (Challis, 1995; Cappello et al., 1996; Spoor and Veldpaus, 1980; Veldpaus; et al., 1988). The transformation is obtained by solving the following optimisation problem:

$$\min \quad f = \sum_{i=1}^{m} (R\mathbf{x}_i + \mathbf{v} - \mathbf{y}_i)^{\mathrm{T}} (R\mathbf{x}_i + \mathbf{v} - \mathbf{y}_i) \tag{1}$$

$$\text{s.t.} \quad R^{\mathrm{T}} R = I, \tag{2}$$

where $\mathbf{x}_i$ and $\mathbf{y}_i$ are position vectors of marker $i$ in the marker array at the reference and current positions, respectively, $R$ is the rotation matrix, $\mathbf{v}$ is the translation vector and $m$ is the number of markers. The constraint $R^{\mathrm{T}}R = I$ ensures that the transformation is orthogonal, corresponding to rigid body motion. The minimum value of $f$ is the segmental residual error ($e$), a measure of the marker array deformation which is mainly due to skin movement artefacts. It is noted that the above formulation was proposed for the kinematics of a rigid body between two subsequent positions with the first position as the reference (Cappello et al., 1996; Challis, 1995; Spoor and Veldpaus, 1980; Veldpaus et al., 1988). The rigid body transformation between the two positions is guaranteed because of Eq. (2) but skin movement artefacts at the reference position were not considered. Chèze et al. (1995) proposed a procedure to define a reference marker array based on all the data frames during motion. However, the selected reference array only represents an average of all the least-deformed arrays during motion. The rigid body elements of the motion of the least-deformed arrays relative to the underlying bone due to skin movement were not taken into account.

Although SOM improves on DM by taking account of skin movement artefacts at the segment level, both methods treat body segments separately without imposing joint constraints (Apkarian et al., 1989; Kadaba et al., 1990; Kepple et al., 1994). Since skin movement patterns in adjacent segments can be very different, errors in poses of the segments will result in apparent joint dislocations or non-anatomical displacements at the joints (Kepple et al., 1994), giving unreliable values of joint kinematics (Genoud, 1996, Lafortune et al., 1992) and uncertainties of joint kinetics. These errors can also have significant effects on the estimation of the lines of action and lever arms of the muscles and the forces transmitted in muscles and other structures.

In this paper, a new method based on the global minimisation of the overall measurement errors with joint constraints for the simultaneous determination of the spatial pose of all segments of a multi-link model of the locomotor musculoskelatal system is presented. It was used to test the hypothesis that consideration of joint constraints and global error compensation can largely reduce the effects of skin movement artefacts on the calculated poses of the musculoskeletal system, particularly the values of axial rotation and ab/adduction at the joints.

## 2. Methodology

The new approach, the *global optimisation method* (*GOM*), is based on the search of an optimal pose of the multi-link model for each data frame such that the overall differences between the measured and model-determined marker coordinates are minimised in a least squares sense, throughout all the body segments. It considers measurement error distributions in the system and provides an error compensation mechanism between body segments, which can be regarded as a global optimisation at the system level.

The pose of an $r$ degree of freedom, $n$-link chain model can be fully described by $r$ generalised co-ordinates $\xi = [\xi_0, \xi_1, \ldots, \xi_r]^{\mathrm{T}}$, such as linear and angular displacements. The model is then customised to individual subjects by using subject specific parameters based on measurements on the subject in a standing position (subject calibration). Since there is no skin movement at this static position, marker arrays are taken as references for subsequent pose estimation during movement. Each marker array is used to define a segment-embedded reference frame and marker position vectors are represented in their local reference frames, denoted together as $P^* = [P_1^*, P_2^*, \ldots, P_n^*]$, where $P_i^* = [\mathbf{p}_1^*, \mathbf{p}_2^*, \ldots, \mathbf{p}_m^*]_i$ are the local marker position vectors on segment $i$.

Given a set of measured marker coordinates $P$ on a data frame, the global optimisation at the system level is to find a set of generalised coordinates $\xi$ such that the following error function

$$f(\xi) = [P - P'(\xi)]^{\mathrm{T}} W [P - P'(\xi)] \tag{3}$$

is minimised where $W$ is a positive-definite weighting matrix. $P'(\xi)$ is the corresponding set of marker coordinates calculated by the following transformation:

$$P'(\xi) = T(\xi)P^*, \quad (4)$$

where $T(\xi)$ is the combined transformation matrix from segment-embedded frames to laboratory frame and is calculated by the model for a given $\xi$. Eq. (4) maintains the integrity of the model because joint constraints are part of the model. It also implies that the segment level rigid body assumption is automatically satisfied. The weighting matrix $W$ is of the following form:

$$W = \begin{bmatrix} W_0 & 0 & 0 & 0 \\ 0 & W_1 & 0 & 0 \\ 0 & 0 & \ddots & \\ 0 & 0 & 0 & W_n \end{bmatrix}, \quad (5)$$

where $W_i$ is a $(3m_i \times 3m_i)$ weighting matrix assigned to the *ith* segment to reflect the error distribution among the $m_i$ markers. For simplicity, in this paper $W_i$ is chosen to give equal weightings to the markers on any one segment. However, each segment is given different weighting factor reflecting its average degree of skin movement artefacts. For this purpose, segmental residual errors are used as a guide so that segments with bigger residual errors should be assigned smaller weightings. For example, skin movement artefacts on the thigh are much bigger than on the pelvis and shank (Cappozzo et al., 1996), with bigger residual error. Therefore, the weighting factor assigned to the thigh must be smaller than those for the pelvis and shank. For each segment, Eqs. (1) and (2) are solved to yield its segmental residual error $e_i$, together with its transformation variables. The weighting matrix $W_i$ is then defined as

$$W_i = \frac{1}{e_i} I. \quad (6)$$

It is noted that different weighting schemes can be used to adapt to specific systems but it will be shown that the proposed scheme produces satisfactory results in simulated gait trials.

The resulting global optimisation problem is a nonlinear programming problem so it has to be evaluated numerically using iterative optimisation methods. Therefore, an initial guess for the design variables $\xi$ is needed. For this purpose, the transformation variables obtained using Eqs. (1) and (2) can be used to derive an initial guess of the model pose.

## 3. Computer experiments

To provide a basis for the comparison of the proposed GOM with other methods, computer experiments were performed based on experimental data from a normal subject. A 3-link chain model of the human pelvis-leg apparatus was developed, with the pelvis, thigh and shank links joined by two ball and socket joints representing the hip and knee respectively. For simplicity, the foot segment was not included in the model in the present study. The model was customised to the subject using marker data measured during a subject calibration trial (Vicon 370, Oxford Metrics, UK). During the trial, markers were placed on key bony landmarks on each body segment and data were captured while the subject was standing upright and stationary. These marker coordinates were used to define segment-embedded reference frames for the associated body segments, following suggestions by Cappozzo et al. (1995). Joint centre positions were then identified and defined in the reference frames of their adjacent body segments. The hip joint centre was identified after Bell et al. (1990). The knee joint centre was taken as the mid-point of the transepicondylar axis and the ankle joint centre as the mid-point of the line joining the two malleoli. Three-dimensional coordinates of the key bony landmarks and joint centre positions were used as the parameters for the customisation of the 3-link model.

Skin movement relative to the bone has been estimated to be as large as 30 mm (Cappozzo et al., 1996). Since there was no skin movement when the subject was stationary during calibration and since the measurement error of the stereophotogrammetric system was estimated to be within 1.3 mm (Gill, 1996), the data obtained enabled an accurate customisation of the model to the subject.

A three-dimensional movement of the 3-link model was generated by applying experimental angular joint kinematics to the model. The positions in space of the marker arrays during motion were calculated on the assumption that the arrays were each rigidly attached to the corresponding segment of the 3-link model. The simulated marker coordinates were determined for 47 frames at 50Hz capturing rate and defined as the true values.

With the true data, 20 computer-simulated gait trials were performed, representing studies of 20 different subjects with different skin movement patterns. For each, a set of movement data was generated by introducing artificial noise into each three-dimensional marker coordinate. A noise model with the form $A\sin(\omega t + \phi)$ (Chèze et al., 1995) was used to simulate skin movement artefacts, where $A$ is the amplitude of the noise, $\omega$ the frequency, $t$ the simulated time and $\phi$ the phase angle. For a given trial, the values of $A$, $\omega$ and $\phi$ for each marker were taken randomly to lie between zero and a set upper limit. The amplitude limit was taken to be 3 cm for the thigh segment and 1 cm for the pelvis and shank (Cappozzo et al. 1996). Since joints extend and flex several times during a single cycle, noise frequencies up to three times that of the measured gait cycle were simulated. Different noise frequencies reflect the combined effects of noise/gait frequency ratios and walking speed. Since skin movement was not zero at heelstrike ($t = 0$), values of $\phi$ up to $2\pi$ were used (Chèze et al., 1995). The model therefore simulated the continuous pattern of the relative skin/bone movement during joint flexion, as observed by Cappozzo et al. (1996).

For the 20 computer trials, joint angles and joint centre positions were calculated using DM, SOM and GOM and compared with the true values. In DM and SOM where the end points of two adjacent segments did not meet, the amount of joint dislocation was calculated as the distance between the two end points, taking the proximal end point of the distal segment as the joint centre. The distal end point of the shank segment was taken as the ankle joint centre simply for comparison of its calculated and true positions. The joint angle convention proposed by Grood and Suntay (1983) was used. Another method which could be used to maintain the integrity of a model and avoid artefactual joint dislocation is to obtain joint angle trajectories calculated by DM or SOM and then apply these data to the model. It may start from the most proximal segment (Top-down) or the most distal segment (Bottom-up). To estimate the likely errors associated with this approach, the joint angles calculated by SOM were applied to the model using a top-down approach (TA). Therefore, there was a total of four methods considered.

## 4. Results

Fig. 1 shows a typical set of joint angles calculated by DM, SOM and GOM, compared with the true values. The ensemble time-averaged errors over the 20 trials were calculated and shown as a bar chart in Fig. 2. Using DM, the average amount of joint dislocation at the hip and knee were 3.88 and 3.24 cm, respectively, (Fig. 3a). The corresponding values for SOM were 1.33 and 0.69 cm (Fig. 3a). Since GOM and TA combined joint constraints, there were no joint dislocations. Although the position of the joint centre of each proximal and distal segment pair is constrained to be coincident in GOM and TA, its position estimated from the noisy data does not necessarily coincide with its true position. The ensemble time-averaged values of the distances between the calculated and true joint positions over the 20 trials were shown in Fig. 3b.

## 5. Discussion

Skin movement artefacts had significant and direct effects on the model poses calculated using DM, as expected. The errors in the calculated joint angles were the biggest among the tested methods (Figs. 1 and 2). The biggest error in the calculated hip joint centre positions was also found in DM (Fig. 3b). SOM took account of

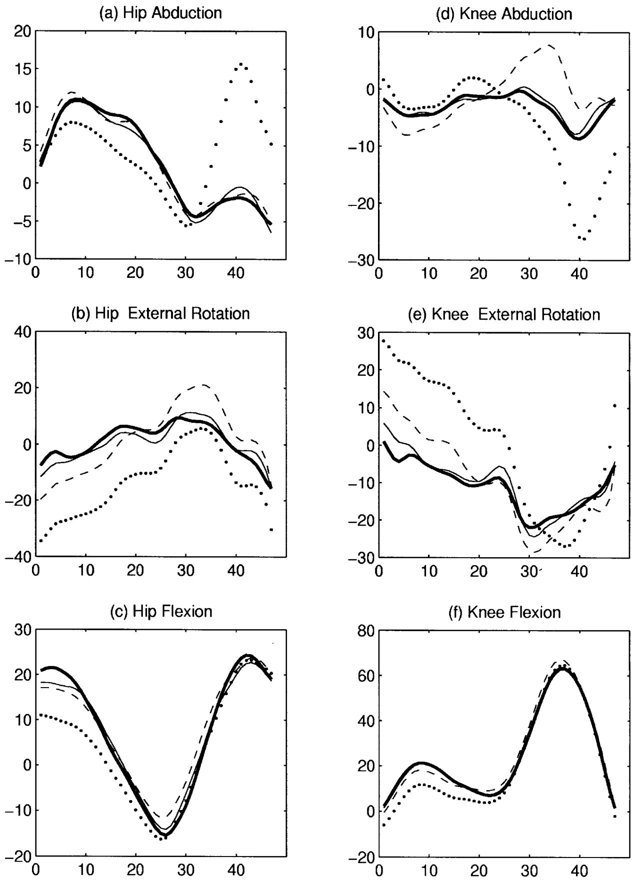


Fig. 1. Results of a typical trial. Joint angles in degrees at the hip (a–c) and knee (d–f) were each calculated using the tested methods (True values: thick solid lines; DM: dotted lines; SOM: dashed lines; GOM: thin solid lines). Horizontal axes are data frame numbers.

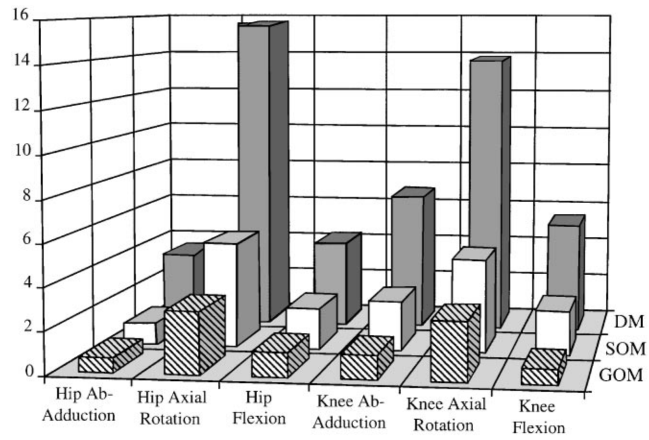


Fig. 2. Ensemble time-averaged errors (in degrees) of the calculated joint angles over the 20 trials.

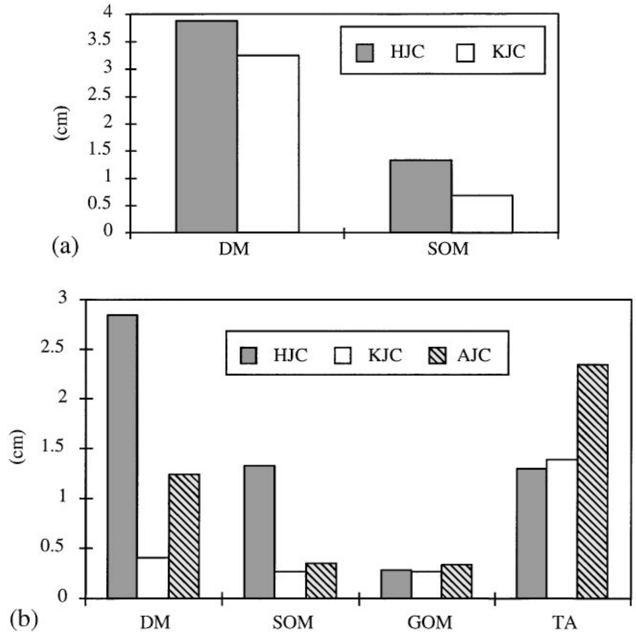


Fig. 3. (a) Ensemble time-averaged values of joint dislocation at the hip and knee for DM and SOM over the 20 trials. (b) Ensemble time-averaged values of the distances between the calculated and true joint positions over the 20 trials (HJC: hip joint centre; KJC: knee joint centre; AJC: ankle joint centre).

skin movement artefacts at segment level so reduced significantly the errors in the calculation. However, relatively big errors were still present in the calculated hip joint positions and joint rotations in the frontal and transverse planes. Like DM, SOM was also liable to joint dislocations (Fig. 3a), meaning the loss of model integrity and violation of the ball-and-socket joint constraints. This indicates that for multi-link systems, optimisation at the segment level does not necessarily guarantee a system level optimum.

With GOM, the problem of joint dislocations was resolved automatically because joint constraints were included in the formulation. Among the tested methods, GOM produced the best results which were very close to the true values (Figs. 1–3). Errors in joint angles of ab/adduction and internal/external rotations were significantly reduced. It is noted that the inclusion of the weighting matrix in the global optimisation formulation together with joint constraints provides an efficient mechanism for error compensation among the body segments. The selection of the weightings according to the amount of the measurement errors allows the estimated pose of a segment with bigger skin movement artefacts to be adjusted towards the true pose by other segments with more accurate measurements. This is clearly shown by the hip joint centre position calculation in Fig. 3b. The more accurate poses of the pelvis and shank helped to bring the thigh back towards the correct pose. Here, joint constraints played an important role in compensating errors. Without global optimisation, however, joint constraints will not compensate but propagate and accumulate errors towards down-stream segments as shown by the joint centre position errors in TA (Fig. 3b). Therefore, incorporation of joint constraints and global optimisation procedures are the two key features of GOM.

It is noted that GOM not only provides a way of imposing joint constraints into skin marker based multi-link models but also takes the full advantage of these joint models in controlling relative motion of body segments. GOM was formulated assuming that each joint of the system is properly modelled and correctly aligned with the adjacent segments. In the present study, the reference motion data were generated by a model with perfect ball and socket joints which were not affected by the applied forces. Therefore, pure kinematic joint constraints were sufficient. For accurate reconstruction of the real musculoskeletal system for the study of its mechanics, more anatomical joint models, other than simple ball and socket joints or hinge joints, are needed. For example, a model reproducing the moving axis of rotation of the natural knee improves the accuracy of the calculated knee joint moment. In patients with abnormal joints, the specific motion characteristics of the joints as a result of the affected structures and applied external and muscle forces should be included. In all cases, correct position and alignment of the joint models are essential. In terms of the selection of weightings, the current weighting scheme worked well in the tested trials but with further consideration of error distributions among the markers even more accurate estimations of the model poses could be achieved. Continuing research on a better alternative weighting scheme is suggested as a topic for further research.

For simplicity, the foot segment was not included in the present model simulation. In studies of living subjects, however, the foot is an important segment for consideration. Markers on the foot are needed. With proper modelling of the ankle joint, the GOM can be applied without difficulty. It also can be applied to the upper limb. However, a limitation of the GOM as well as any other methods which are based on the assumption of rigid body segments is their application to the trunk segment. Further research effort is needed for the establishment of an appropriate representation of the trunk in gait analysis.

The GOM has been shown to provide an efficient and reliable method for the calculation of the poses of multi-link models from marker coordinates. When implemented on a PC with a 3D computer graphics-based locomotor system model (Lu and O'Connor, 1998), it requires less than a minute to analyse a gait trial to produce all the gait variables and data for 3D model animation. The GOM removed the possibility of joint dislocation. With more anatomical joint models, the method may be used to reduce the effects of measurement errors, predominantly skin movement artefacts, on the reconstructed poses of the musculoskeletal system and its subsequent mechanical analyses. From the present comparative study, it was shown that consideration of joint constraints and global error compensation can largely reduce the effects of skin movement artefacts on the calculated poses of the musculoskeletal system. Methods based on a concept similar to that of GOM may be useful in clinical gait analysis and in computer graphics based model animation where realistic motion is essential and non-anatomical joint dislocations are not allowed. GOM minimises errors in axial rotation and ab/adduction at the joints and may extend the applicability of gait analysis to clinical problems.

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