Kinematic Reconstruction of the Lower Limb Based on Measurements of the Body Surface

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Abstract— **This paper focusses on a knee model to support 4D (3D+time) optical measurements of the body surface. The use of optical techniques such as rasterstereography in gait analysis is in full development and will be an alternative for the state-of-the-art techniques based on tracking external markers. The extensive information in the acquired surface offers the possibility to describe the motion more accurately as compared to the description based on tracking external markers. However, the currently used knee model, a 1dof hinge joint, limits the use of the extensive information since it can not explain the real knee movement. Furthermore, as every person is different an accurate model needs to be individualised.**

This paper presents a procedure to individualise a four-bar knee model and hence to use it to describe the kinematics of a measured motion. Determining the kinematic parameters of the individualised four-bar constitutes a non-linear optimisation problem which suffers from local minima. However, using a circle involute model to provide a so called 'hot start' results in a 'good enough' local optimum. By linking the circle involute model and the four-bar model, individualised kinematic parameters can be estimated based on measurements of the moving body surfaces.

Three measurements validate the concept.

Keywords— **biomechanics, individualised model, four-bar**

I. Problem Statement

In clinical practice, motion analysis is an indispensable tool with regard to the treatment of gait pathologies, rehabilitation and the improvement of prosthesis design. Other important applications of motion analysis include ergonomics and sports.

A motion analysis starts with the measurement of the motion. In a second step, kinematic data is extracted from these measurements. The importance of an accurate kinematic description is twofold: the kinematic data contains important clinical data and is at the base of dynamic calculations.

The state-of-the-art method to measure human motion consists of tracking external markers fixed on the skin of the patient, measuring the ground reaction forces and recording the electromyograms of some superficial muscles ([1], [2]). Video cameras record the positions of the active or passive markers during the motion. Active markers e.g. emit light, passive markers reflect the incoming light. A fundamental step in this approach is the determination of the relation between the markers fixed on the skin and the underlying bones. This is done in a calibration procedure that usually consists in measuring the markers and the underlying bony structures simultaneously in a stationary patient. This results in a coordinate system associated with the markers and a coordinate system associated with the underlying bone for each body segment. In processing the measurement data the assumption is made that the relation between these two coordinate systems is fixed, in other words that each body segment is a rigid body. However, in reality the skin and consequently the markers fixed to the skin move with respect to the underlying bones. The resulting errors are called motion artifacts. These motion artifacts are kept down to a minimum by accurately attaching the markers where the relative motion of the skin with respect to the underlying bone is smallest. As a consequence, attaching the markers is a labour-intensive procedure that takes, together with the calibration, between half an hour and an hour. Moreover a specialist has to carry out this procedure. This long measurement procedure not only contributes to the high staff cost but also results in the use of a limited amount of markers where it comes to children in order to avoid difficult motion analysis.

The use of optical techniques such as rasterstereography ([3], [4]) in motion analysis is in full development and will be an alternative for the state-of-the-art techniques based on tracking external markers. Rasterstereography is an optical technique to measure surfaces. The technique is based on the projection of a regular light pattern on the object to be measured. The distortion of the light pattern depends on the curvature of the surface. A video camera records the distorted pattern. The calculation of the spatial coordinates of the points on the surface is based on the digital camera images and triangulation. This results in a set of points, describing the measured surface. The curvature of the surface in these points can be calculated. Rasterstereography is already common practice in the treatment of scoliosis [3].

The use of rasterstereography for motion analysis has several important advantages as compared to the systems based on tracking external markers. Rasterstereography does not use markers but measures surfaces. Anatomical landmarks are bony structures, recognizable from the measured surfaces because of their superficial location. Specific values of the curvature of the measured surfaces characterise these anatomical landmarks and allow to retrieve their position. Eliminating the use of markers saves on time as well as on costs. In contrast to the markers which move along with the skin, these curvatures have a direct relation to the underlying bony structures. Therefore rasterstereography is implicitly less sensitive to motion artifacts.

The body surface acquired by optical systems contains different and more extensive information compared to the limited information provided by the markers. The use of the surfaces is not limited to detecting the position of the anatomical landmarks. For instance, based on the acquired surface the muscle mass can be estimated, the segment-axes can be determined and larger anatomical structures such as the patella and the margo anterior of the tibia can be detected.

In summary, motion analysis based on measurements of the surface avoids the use of markers, is less sensitive to motion artifacts and contains more extensive information in comparison to the common systems. Because of this, it has the potential of describing the motion more accurately based on a less time-consuming measurement procedure.

In the state-of-the-art motion analysis, relevant clinical data is extracted from the measurements by fitting a model to them. In this way the model filters the data which results in a more accurate description of the measured motion. In motion analysis, based on measurements of the body surface, a model can be used in a similar way. However, adapted models are needed for two reasons. On the one hand the information contained in the measurements is different. On the other hand the more extensive information in the acquired surfaces and the smaller sensitivity to motion artifacts offer the possibility to describe the motion more accurately. Especially the most commonly used knee model, a 1dof hinge joint, is not sufficient since it can not explain the real knee motion which consists of a combination of rolling and sliding. Furthermore, as every person is different, an accurate model needs to be individualised.

The present paper makes a double contribution. First, it presents a procedure to individualise a four-bar knee model based on measurements of the body surface and second, a procedure to use the model to extract relevant clinical data.

II. Kinematic Musculoskeletal Modelling

This section gives a brief overview of the kinematic knee models described in literature: the hinge joint, the Euler angle description, the centroids description, the helical axis method, four fixed rotation axis description, the four-bar model and the circle involute model (see also [5]).

The simplest model of the knee is a hinge joint. However, the simple hinge joint gives only an approximate description of the real knee movement which is a combination of rolling and sliding.

The Euler angle description [6] describes the kinematics of the knee by the relative translation and rotation of the femur with respect to the tibia. Subsequent rotations, expressed by e.g. Euler angles, describe this relative rotation. The advantage of this method is that the coordinate system can be chosen as to obtain a translation component and rotation angles related to clinical terms. However, no standard is available for the choice of the coordinate system which makes comparison between different systems difficult. Strictly speaking the Euler angle description is not a knee model but a method to describe the knee motion.

Centroids desribe the path of the instantaneous rotation center of the knee. Freudenstein and Woo [7] approximate the centroid, which is the path of the intantaneous rotation center of the knee, by a simple curve, such as a logarithmic spiral, while reproducing the knee motion with reasonable accuracy. Since the centroids are motion dependent it is difficult to apply this method to a general knee motion.

The helical axis method is based on a well known mechanical principle: the three-dimensional motion of two rigid bodies with respect to each other can be described as a combination of a rotation around an axis and a translation along the same axis. In literature this axis is called helical axis, twist axis, screw axis or instantaneous axis of rotation. For a planar motion the helical axis reduces to the instantaneous center of rotation. The method was first adopted by Blacharski et al. [8] and later on by many others [5]. In this method, the helical axis or the instantaneous center of rotation is estimated during different intervals of the motion. This method allows for a description of the motion that is independent of the choice of coordinate system. However, estimation of the axis is difficult since the method is sensitive to measurement errors.

A recent development is the description of the knee motion using four fixed independent rotation axes [9]. At each time instant the motion of the knee can be described as a combination of rotations around three fixed axis. However, the location of one of these axis is dependent on the flexion.

The description of the movement of the knee in the sagittal plane using a four-bar (Figure 1) has a long history and is, starting from '70, well accepted ([10], [11], [12], [13], [14], [15]). The crossed bars in this mechanism represent the cruciate ligaments which have approximately a fixed length, since during the largest part of the knee motion it is possible to find a nearly isometrical fibre [16]. In the literature the four-bar is often used to describe the knee motion since it is based on the anatomical structures of the knee. The mechanism gives a good description of the knee motion in the sagittal plane which is a combination of rolling and sliding. Furthermore, the joint surfaces extracted from this mechanism are a good approximaton of the real joint surfaces and the measured contact points between tibia and femur nearly coincide with those measured from cadavers [15]. The individualisation of the parameters of the four-bar is until this moment limited to linking the lengths of the four bars and the location of implant to the tibia and femur to measurements on cadaver knees. The four-bar in the sagittal plane can be extended to a three-dimensional bar model [17].

Rehder [18] showed, using measurements on cadaver knees, that a circle involute (Figure 1) accurately describes the contact surface of the femur in the sagittal plane based on the measurements on cadaver knees. Bao and Willems [5] developed a procedure to determine the circle involute and to estimate the contact surfaces based on measurements of the position of markers on tibia and femur.

The knee model used in this paper is the four-bar model. The contribution of this paper is the development of a procedure to individualise this model based on measurements of the kinematics of a calibration motion. The individualisation of the four-bar mechanism makes use of the circle involute model as explained in the next section.

III. Individualisation of the four-bar mechanism

The aim of the individualisation is to determine the model parameters based on a measurement of the kinematics. The model parameters are the kinematic parameters of the four-bar and the position of the anatomical landmarks relative to this four-bar. The modelled anatomical landmarks

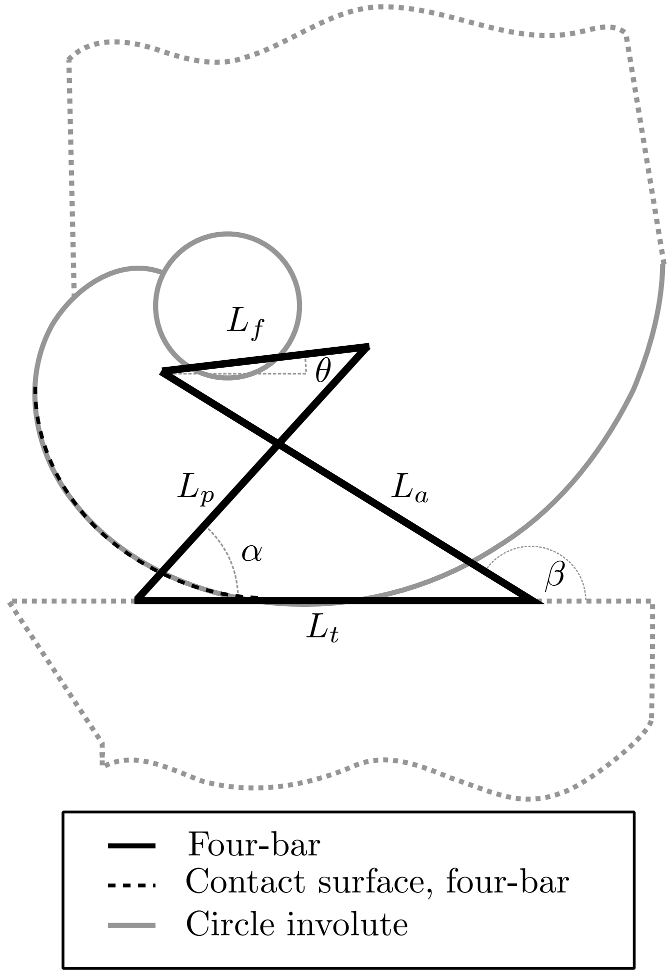

Fig. 1. Knee model

are fixed with respect to the four bar while this is, due to the measurement errors, not the case for the measured anatomical landmarks. Determining the model parameters is formulated as an optimisation problem. The objective function penalises the difference between the measured position of the anatomical landmarks:

$$\begin{aligned} \tilde{x}_{ji} \qquad i &= 1 \ldots N \\ j &= 1 \ldots n_l + n_k \ , \end{aligned} \tag{1}$$

and the modelled position of these landmarks:

$$\begin{aligned} x_{ji} \qquad i &= 1 \ldots N \\ j &= 1 \ldots n_l + n_k \ . \end{aligned} \tag{2}$$

The index i indicates the different time instants and N denotes the total number of considered time instants. The index j indicates the different anatomical landmarks, n_l denotes the total number of anatomical landmarks on the lower leg and n_t denotes the total number of anatomical landmarks on the thigh. Since the four-bar model is a 2D-model, the objective function is expressed in the motion plane of the knee. The resulting optimisation problem is nonlinear and suffers from local minima. Therefore, the circle involute model is used to provide a hot start, that is an initial value for the optimisation variable that is known to be a good approximation of the optimisation variable in the optimum. This results in a 'good enough' local optimum.

Figure 2 schematically outlines the procedure for the individualisation of the four-bar mechanism.

A. Projection of the data

At each measured time instant, the result of a raster-stereographic measurement is a set of points, describing the body surface. Based on the curvature of the surface in these points, the position of the anatomical landmarks is detected. This measured position is characterised by its coordinates in a world coordinate system that is fixed with respect to the environment. Since the optimisation problem for determining the model parameters is expressed in the motion plane of the knee, the 3D-positions of the anatomical landmarks need to be projected in this plane.

When the knee moves, the relative motion of the thigh with respect to the lower leg is largely in a plane, the motion plane of the knee. This relative motion is easily described in a coordinate system attached to the lower leg, called the lower leg coordinate system. In such a coordinate system the anatomical landmarks on the lower leg stand still and the motion of the anatomical landmarks on the thigh is the relative motion with respect to the lower leg. Due to the measurement errors, there exist no coordinate system in which the measured anatomical landmarks stand still. Therefore the problem of determining the lower leg coordinate system is formulated as an optimisation problem. The objective function penalises the motion of the anatomical landmarks on the lower leg in the lower leg coordinate system. This objective function is minimised under the constraint that the motion of the lower leg is continuous. The optimal variables determine the pose, which is position and orientation, of the lower leg coordinate system with respect to the world coordinate system at each considered time instant and the position of the modelled anatomical landmarks on the lower leg:

$$^{l}x = [^{l}x_1 \ldots {}^{l}x_N] \tag{3}$$

$$\text{with} \quad ^{l}x_i = [x_{1i} \ldots x_{n_l i}] \ . \tag{4}$$

It is then straightforward to transform the anatomical landmarks on the thigh and the lower leg to the lower leg coordinate system.

In the lower leg coordinate system, the anatomical landmarks on the thigh approximately move in parallel planes. The common normal of these planes determines the motion plane of the knee. The problem of finding this normal is again an optimisation problem in which the component of motion out of these planes is minimised. Once this normal is known, the position of the landmarks can be projected on the motion plane of the knee. This projection can be done both in the lower leg coordinate system and in the world coordinate system. A projection in the lower leg coordinate system facilitates the formulation of the model parameter determination.

The choice of the lower leg as reference was not arbitrary. Since the lower leg contains more bony structures with

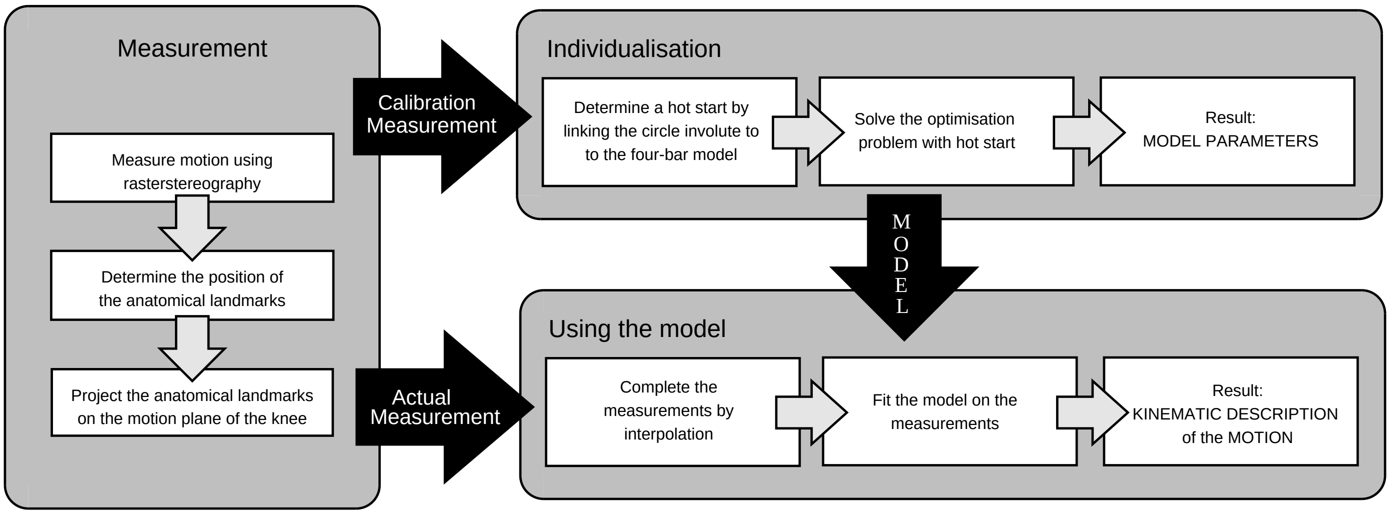

Fig. 2. Procedure for individualisation and using the model

a superficial location than the thigh, a lower leg coordinate system can be determined with higher accuracy than a thigh coordinate system.

B. Providing a hot start

Bao and Willems [5] describe a procedure to estimate the parameters of a circle involute, describing the contact surface of the femur, based on a kinematic measurement of the knee motion. They describe how to use the estimated circle involute to determine the contact points on both the femur and the tibia for each measured time instant. The procedure of Bao and Willems is used here with one extension. The normal on the tibia plateau is not directly measured but is determined by an optimisation. The normal, resulting in the smallest residual for the objective function of Bao and Willems, is used.

An important remark has to be made on the circle involute model. Describing the contact surfaces removes only one out of the three dofs that two moving bodies in a plane have with respect to each other. This means that given the flexion angle and the contact surfaces, the position of the knee can not be determined unambiguously. In contrast the four-bar links the position of the lower leg to the position of the thigh leaving only a single degree of freedom (dof), i.e. the flexion angle.

The kinematic parameters of the four-bar are the lengths of its bars, $L = [L_t \quad L_f \quad L_a \quad L_p]$ and its pose with respect to the lower leg, ${}^m_l p$ where l refers to the lower leg. A first estimate of these parameters, denoted with $\hat{}$, is determined imposing that the contact points on the tibia and femur following from the estimated four-bar coincide with the contact points calculated using the circle involute model. This constitutes solving an overdetermined set of equations which is done in a least square sense. Along with the estimate of the kinematic parameters an estimate of the configuration of the four-bar, described by the flexion angle, $\theta = [\theta_1 \dots \theta_N]$, is obtained.

There is a good agreement between the original contact surface and the contact surface that follows from the four-bar (Figure 1).

C. Determining the model parameters

The optimisation problem to determine the model parameters has the following structure:

Optimisation variable

The optimisation variable is:

$$z = [L \quad {}^m_l p \quad {}^t x \quad \theta \quad \alpha \quad \beta] \ , \tag{5}$$

with

$$ {}^l x = [{}^t x_1 \dots {}^t x_N] \tag{6}$$

$$\text{with} \quad {}^t x_i = [x_{(n_l+1)i} \dots x_{(n_l+n_t)i}] \ , \tag{7}$$

where $\alpha = [\alpha_1 \dots \alpha_N]$ and $\beta = [\beta_1 \dots \beta_N]$ are defined as in Figure 1. Since the four-bar has only 1dof, α and β follow unambiguously from θ. This relation is expressed by the loop closure equations of the four-bar. Since α and β follow from θ, they are not strictly necessary to formulate the optimisation problem. However, introducing these two vectors of extra optimisation variables together with the constraints that describe the dependency, has a positive influence on the convergention speed of the optimisation problem.

Objective function

The objective function is:

$$f(z) = \sum_{i=1}^{N} \Delta^t X_i^T W_t \Delta^t X_i \ , \tag{8}$$

where ${}^t X_i \in \mathbb{R}^{\,2n_t}$, the difference between the modelled, ${}^t x_i$, and measured position, ${}^t \tilde{x}_i$, of the anatomical landmarks on the thigh at time instant i, is defined as

$$\Delta^t X_i = [{}^t x_i - {}^t \tilde{x}_i] \ . \tag{9}$$

W_t is a weight matrix that allows to differentiate between the different landmarks to take into account the reliability of the measurement of the position of the considered landmark.

Constraints
The loop closure equations express the relation between θ, α and β.

Hot start
The hot start:

$$z_0 = [\hat{L} \quad {}^m_l\hat{p} \quad {}^t\tilde{x} \quad \hat{\theta} \quad \hat{\alpha} \quad \hat{\beta}] \ , \tag{10}$$

is obtained as described in the previous paragraph while $\hat{\alpha}$ and $\hat{\beta}$ are obtained from $\hat{\theta}$ by applying the loop closure equations.

IV. Using the model

The model is used to determine the position of the joints and the joint angles during the measured motion. This is done by fitting the model on the measurements. The result of this fit is the position and configuration of the four-bar which is directly related to the flexion angle of the knee. Fitting the model on the measurements constitutes a nonlinear optimisation problem for each considered time instant. The objective function penalises the difference between the measured position of the anatomical landmarks and the modelled position of these landmarks. Good starting values for the positions of the landmarks are needed for a fast convergence. These starting values follow from fitting the segments separately.

Figure 2 schematically outlines the procedure for using the model.

A. Fitting of the model on the measurements

Fitting of the model is possible if sufficient landmarks are measured during the movement: three landmarks, at least one on each segment, are needed. The aim of the fitting is to determine the flexion angle and the pose of the four-bar with respect to the world, ${}_wp_i$ where w refers to the world coordinate system, at each time step. The model, with the individualised model parameters, is used in this procedure. To fit the model on the measured positions of the landmarks a least square objective function is formulated which is a measure for the fit between the modelled position of the landmarks and the measured position of the landmarks. The optimisation problem to determine the pose and configuration of the four-bar has the following structure:

optimisation variable
The optimisation variable is:

$$z_i = [{}_wp_i \quad \theta_i \quad \alpha_i \quad \beta_i] \ . \tag{11}$$

Objective function
The objective function is:

$$f(z_i) = \Delta^l X_i^T W_t \Delta^l X_i + \Delta^t X_i^T W_t \Delta^t X_i \ , \tag{12}$$

where the definition of $\Delta^l X_i$ and W_l is analogous to the definition of $\Delta^t X_i$ and W_t.

Constraints
The loop closure equations express the relation between θ_i, α_i and β_i.

Hot start
The following paragraph describes how the hot start:

$$z_{0i} = [{}_w\hat{p}_i \quad \hat{\theta}_i \quad \hat{\alpha}_i \quad \hat{\beta}_i] \ , \tag{13}$$

is obtained.

Every time instant the nonlinear optimisation problem with nonlinear constraints has to be solved. This optimisation problem needs a hot start to guarantee the convergence to a 'good enough' local optimum. Moreover, a hot start accelerates the convergence.

B. Providing a hot start

The aim is to find, in a numerically inexpensive and robust way, an estimate for the pose of the four-bar with respect to the world, ${}_w\hat{p}_i$, and for the configuration of the four-bar, $\hat{\theta}_i$. These estimates are used as the hot start for the optimisation problem described in the previous paragraph. These estimates follow from a separate localisation of the lower leg and the thigh and the use of the individualised model.

At each considered time instant the modelled positions of the anatomical landmarks on the lower leg can be fitted on the measured positions of these landmarks by minimising the following objective function:

$$f({}^l\hat{p}_i) = \Delta^l X_i^T W_l \Delta^l X_i \ . \tag{14}$$

The optimisation variable, ${}^l\hat{p}_i$ is a good estimate of the pose of the lower leg. This estimate is obtained efficiently. First, a minimal value for the objective function is obtained in a few iterations. Second, the optimisation problem is robust, convergence to the same optimum is observed with substantially different initial values. Since the pose of the four-bar with respect to the lower leg is known from the individualised model, an estimate of the pose of the four-bar with respect to the world, ${}_w\hat{p}_i$ is easily obtained from ${}^t\hat{p}_i$.

An estimate of the pose of the thigh, ${}^t\hat{p}_i$, follows analogously from a fit of the modelled positions of the anatomical landmarks on the thigh to the measured positions of these landmarks. The pose of the lower leg and the pose of the thigh determine the flexion angle. This gives an estimate of the configuration of the four-bar $\hat{\theta}_i$ based on ${}^l\hat{p}_i$ and ${}^t\hat{p}_i$.

$\hat{\alpha}_i$ and $\hat{\beta}_i$ are obtained from $\hat{\theta}_i$ by applying the loop closure equations.

C. Interpolation

Interpolation is necessary when the surface images temporarily do not provide sufficient information. The interpolation should make optimal use of the data which is still available.

V. Experimental Data and Results

A. Experimental Data

The procedures proposed in this paper are validated using three measurements on three different normal subjects. The data is acquired with the state-of-the-art techniques based on markers attached to the skin, since the use of rasterstereography to record moving surfaces is not yet well established. The motion of the first subject is measured using the Vicon system with six passive markers attached to the lower leg and two passive markers attached to the thigh. The motion of the other two subjects is measured using a Krypton (Metris) system with four active markers both on the lower leg and thigh.

Due to the use of a marker-based technique, skin artifacts corrupt the measurements. The maxima of these cyclic artifacts are listed in Figure 3(a) for all subjects and for both lower leg and thigh. This artifacts should be kept in mind when evaluating the results.

B. Results

Individualisation

The described procedure for individualisation results in an individualised knee model for all subjects. The obtained bar lengths are listed in Figure 3(b).

The distance between the positions of the anatomical landmarks as depicted by the model and the measured anatomical landmarks is a measure for the quality of the individualised model. These distances are listed in Figure 3(c). Taking into account the skin artifacts (Figure 3(a)), the results are very satisfying.

The entire procedure is implemented in Matlab. The processing time needed for individualisation is low, the individualisation of the model is nearly online on a normal PC. Furthermore, the calculation time is neglegible compared to the time needed for acquisition and reconstruction of the surfaced and the calculations of the curvatures of the surface, which are inherent to rasterstereographic measurements.

Using the model

The described procedure for using the model is able to fit the individualised model on the measured movements, different from the calibration movements, for all subjects.

The distance between the positions of the anatomical landmarks as depicted by the model and the measured anatomical landmarks is a measure for the quality of the fit of the individualised model to the measured motion. These distances are listed in Figure 3(d). Taking into account the skin artifacts (Figure 3(a)), the results are again very satisfying.

The entire procedure is implemented in Matlab. The model can be used quasi online. The calculation time is small compared to the individualisation time.

VI. Conclusions

This paper introduces a procedure for individualising a kinematic model of the knee based on measurements of the body surface. The proposed model is a four-bar mechanism since such a mechanism can describe a combination of rolling and sliding, as opposed to the commonly used hinge joint.

The problem of determining the kinematic parameters of the four-bar is solved efficiently using the circle involute model to provide a hot start for the original nonlinear optimisation problem. The circle involute model is used to estimate contact surfaces that explain the measured motion. A first estimate of the kinematic parameters of the individualised four-bar is based on these contact surfaces. This first estimate is the hot start for the original optimisation problem which results in the parameters of the individualised four-bar. By linking the circle involute model and the four-bar model, individualised kinematic parameters can be estimated based on measurements of the moving body surfaces.

This paper also introduces a procedure for using the model to extract relevant clinical data from the measurements.

The applicability of these procedures is demonstrated by applying them to three data sets, acquired with the state-of-the-art techniques, since the optical techniques are not yet well established. It is shown that (i) an individualised model and a fit are found with a processing time that is negligible compared to the processing time needed for the acquisition and reconstruction of the surfaces; (ii) the fit between the individualised model and the measurements is in the order of magnitude of the measurement accuracy.

Further work is needed to demonstrate the full power of the developed procedures. In measurements of the body surface, the quality of fit between the model and the measurements is no longer limited to the relatively low measurement accuracy of the ordinary techniques. It is necessary to use the method in combination with the optical techniques for measuring the body surface to show the advantage of the individualised four-bar in comparison to the hinge joint. In addition, the implementation of a Kalman-filter to fit the model to the measurements offers the possibility to use the information contained in the measured surface more efficiently.

Acknowledgments

Friedl De Groote and Tinne De Laet are research assistants of the Fund for Scientific Research (FWO) - Flanders (Belgium).

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max artifact [*mm*]	Lower Leg	Thigh
subject 1	19	23
subject 2	14	8
subject 3	8	12

(a) Skin artifacts

bar lengths [*mm*]	L_a	L_p	L_t	L_f
subject 1	98.46	52.28	62.00	34.73
subject 2	43.95	23.96	31.87	11.56
subject 3	64.53	73.36	19.31	3.13

(b) Bar lengths

max distance [*mm*]	Lower Leg	Thigh
subject 1	5	5
subject 2	10	5
subject 3	4	6

(c) Distance between modelled and measured landmarks, individualisation

max distance [*mm*]	Lower Leg	Thigh
subject 1	5	9
subject 2	11	7
subject 3	7	8

(d) Distance between modelled and measured landmarks, using the model

Fig. 3. Experimental results

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