A computational model of postoperative knee kinematics

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Abstract

A mathematical model for studying the passive kinematics of total knee prostheses can be useful in computer-aided planning and guidance of total joint replacement. If the insertion location and neutral length of knee ligaments is known, the passive kinematics of the knee can be calculated by minimizing the strain energy stored in the ligaments at any angular configuration of the knee. Insertions may be found intraoperatively, or may come from preoperative 3D medical images. The model considered here takes into consideration the geometry of the prosthesis and patient-specific information. This model can be used to study the kinematics of the knee joint of a patient after total joint replacement. The model may be useful in preoperative planning, computer-aided intraoperative guidance, and the design of new prosthetic joints. © 2001 Elsevier Science B.V. All rights reserved.

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1. Introduction

One of the major goals of total knee replacement is to restore the normal function of the knee. The success of restoring a patient’s knee function depends, among other factors, on the design and implantation placement of the prosthesis. The function of the knee after total joint replacement is heavily influenced by the design of the prosthesis and the surgical placement of the prosthetic devices (Garg and Walker, 1990). This work presents a computational technique for understanding knee motion, with a primary goal of producing improved computer-aided implantation of artificial knee joints.

A set of knee-prosthesis components consists of: an anatomically shaped distal femoral element, normally made of a cobalt-based alloy; a proximal tibial element that is normally made of ultra-high-molecular-weight polyethylene (hereafter, simply “polyethylene”); and an optional patellar insert, typically made of polyethylene. Implantation of a prosthesis typically requires removal of the anterior cruciate ligament (ACL) and, depending on the prosthesis design, may also involve removal of the posterior cruciate ligament (PCL). The medial collateral ligament (MCL) and lateral collateral ligament (LCL) are critical in holding the joint in place and producing joint motion. Fig. 1 shows the knee before and after implantation of the prosthetic components.

In passive knee motion, where no external forces are present, the femur is kept in contact with the tibia by the tensile forces exerted by the surrounding soft tissues. The geometry of the articular surfaces provides a set of feasible contact locations that determine the orientation of the knee. The interaction of the surrounding ligaments thus governs the contact conditions of the knee. It is supposed here that, at any given angulation, the contact condition that the knee would naturally assume is the contact condition that would minimize the total strain energy stored in the ligaments of the knee.

The theoretical goal of this work was to understand interactions between the geometry of the articular surfaces and the surrounding ligaments and to study how such interactions affect the overall kinematics of the knee. The passive kinematics were analyzed as the instantaneous quasi-static solution to ligament strain-energy minimiza-
tion. This model required knowledge of the geometry of the articular surfaces, the insertion locations and mechanical properties of the knee ligaments, and the implanted location of the prosthetic components. The model determined the contact location between the femoral and tibial articular surfaces that minimized the strain energy stored in the ligaments of the knee. The ligament states, contact trajectories, and passive knee kinematics were analyzed to predict the postoperative kinematics. The knee was also animated, using standard computer graphics techniques, to provide a visualization of the passive motion of the artificial knee components.

1.1. Previous work

It is well known that the geometry of the articular surfaces can affect the location of the contact point during knee motion and ultimately affect the trajectory of the leg (Kurosawa et al., 1985; Essinger et al., 1989). The position and orientation of the components can also significantly affect the pattern of knee motion (Martelli et al., 1998) because of the effects of the ligaments during knee motion (Kapandji, 1967; Lew and Lewis, 1978; Butler et al., 1986; Bradley et al., 1988). Surgical implantation strategies, such as placement and sloping of the components, can have dramatic effects on gait (Garg and Walker, 1990). All these factors affect the knee kinematics in a complex manner such that changing a single factor often results in a global change of the knee kinematics.

The mathematical model presented by Essinger et al. (1989) introduced the concept of system-energy minimization for use in determining the kinematic state of the knee. The model required as input the attachment coordinates and mechanical parameters of the ligaments, as well as the 3D geometrical description of the articular surfaces (prosthetic or natural). The model generated the kinematics of the joint, the motion of the center of contact, the quadriceps forces, the pressure distribution on the tibial plateau, and the ligament lengths and forces between 0° and 120° of flexion. When applied to the natural “averaged” knee, the model predictions were reported to be consistent with those reported in the literature. When applied to three different commercially available prosthetic designs, the model revealed that condylar-type prosthetic knee motions were closely related to the geometry of the bearing surfaces.

The three-dimensional model proposed by Blankevoort’s group (Blankevoort et al., 1991; Blankevoort and Huiskes, 1996; Mommersteeg et al., 1996) is one of the most complex knee models in the current literature. It required as input the geometry of the articular surfaces and the ligament insertions, plus the mechanical properties of the ligaments and the cartilage (for natural knees) or the polyethylene tibial component (for prosthetic knees). The kinematics of the knee were solved by equilibrating the forces and moments on and about the soft tissues and deformable surfaces. The study revealed that the inclusion of deformable contact did not substantially alter the motion.
characteristics from that predicted by non-deformable contact (Blankevoort et al., 1991). The work also introduced the idea of describing the contact of two articular surfaces by means of a set of governing equations. The authors advocated the use of multi-fiber ligament bundles that had non-uniform mechanical properties and neutral-force lengths (Mommersteeg et al., 1996), all of which features have been included in the model presented here.

The 3D mathematical model of the knee presented here is an extension of our 2D model (Martelli et al., 1998) of the planar sagittal motion of a prosthetic knee. The previous model relied on knowledge of the prosthetic bearing surface (derived from a manufacturer’s specifications) and intraoperative measurements of ligament data. Based on the principle of ligament-strain minimization, the model determined the quasi-static knee kinematics by finding successive contact points on the femoral and tibial bearing surfaces over a range of flexion angles. The knee model was fast to compute and demonstrated a plausible sagittal-plane motion. The results of the knee-kinematics model were consistent with several assumptions that are crucial to the development of a model of the human knee:

- the principle of ligament-strain minimization can serve as a useful constraint for knee motion;
- knee kinematics are affected both by the geometry of articulating surfaces and by the surrounding ligaments; and
- the use of a multi-fiber ligament model can produce realistic knee motion.

This model was also generic, in the sense that it could be applied to other body joints with either prosthetic or natural articular surfaces. The motion studied was passive, which means no external forces are considered and articular surfaces were assumed to be rigid. However, extension of this model to study the dynamics of the joint and the inclusion of deformable articular surfaces is also possible.

2. Methods

Many current knee prostheses are designed to have surfaces that are not geometrically congruent (O’Connor and Goodfellow, 1991) for reasons related to human biomechanics and the wear properties of the materials that are used in the prosthetic components, although semi-congruent and fully congruent designs have been and are still used. For example, looking at a slice of a non-congruent knee component through the sagittal plane, the location of idealized point contact has an additional degree of freedom. The femoral component, in a given flexion angle, can be placed in contact with the tibial component anywhere along the bearing surface of the tibial component. Fig. 2 shows cross-sections of a component and how contact might occur.

In our knee model, we include patient-specific ligament data to provide the necessary constraints for describing passive kinematics. Each ligament filament is modeled as having a particular length/strain relation, so passive knee kinematics can be computed by finding the contact state that minimizes the total energy in the ligaments surrounding the knee. That is, each contact state will stretch the ligaments and produce internal strain energy in the filaments; the contact state that minimizes strain energy is the local equilibrium to which the knee components would relax if disturbed from this equilibrium position.

Passive knee kinematics is described as a series of instantaneous quasi-static solutions to energy minimization of a system in which contact is ideal single-point contact. Potential energy stored in a passive knee consists of those stored in each filament of ligaments only, since no external load is present. Friction between the articular surfaces is assumed to be negligible. Both metallic femur and polyethylene tibia are assumed to be rigid.

2.1. Relative joint position and kinematic constraints

The coordinate systems used in this model were similar to commonly used systems (Blankevoort et al., 1991; Martelli et al., 1998). Two Cartesian coordinate systems were assigned to the major bones of the lower limb. The absolute, space-fixed coordinate system was associated with the tibia, while the relative, body-fixed coordinate system was associated with the femur (Fig. 3). The Z axes were aligned with the anatomical axis of the limb, with the proximal direction being positive. The X axes were perpendicular to Z, lying in the sagittal plane with the anterior direction being positive. The Y axes were derived as \( Y = Z \times X \). Without loss of generality, the origin of the absolute coordinate system was located on the mid-point of the tibial component in the medial-lateral direction, lying
on the resection plane that the tibial component was fixed on; the origin of the relative coordinate system was located on the distal cut for the femoral component, also on the mid-point of the femoral component in the medial-lateral direction. The pose of each prosthetic component was specified with respect to its associated coordinate system.

Coordinate systems are related by rigid-body transformations. If \( \mathbf{p} \) is the vector that measures the coordinates corresponding unit normals \( \mathbf{n}_m, \mathbf{n}_l, \mathbf{n}_m, \mathbf{n}_l \) for which there was some transformation \( T(\cdot) \) such that

\[
\hat{C}_m = R(C_m) + \hat{d},
\]

\[
\hat{C}_l = R(C_l) + \hat{d},
\]

\[
\hat{n}_m = -R(n_m),
\]

\[
\hat{n}_l = -R(n_l).
\]

These equations specify that ideal contact occurs when surface points coincide in space, with point normals pointing in opposite directions.

The contact 4-tuple was the primary variable in the preoperative computation. For each given prosthesis orientation, ranging from full extension to deep flexion and through the desired range of internal/external rotation, a set of feasible contact locations was determined as a set of contact 4-tuples, i.e., as a pair of matching point pairs. A match was found for the medial compartment if a femoral point and tibial point could be translated, and rotated in the varus/valgus direction, so that the normals matched within the comparison tolerance. If, for the same varus/valgus angulation, there was a match in the lateral compartment then the matches were determined to be a contact 4-tuple. The match was found by exhaustive search, but heuristics can doubtless be developed to improve on the exhaustive algorithm. Fig. 4 illustrates a contact 4-tuple.

A contact 4-tuple established a varus/valgus angulation of the joint. This angulation, plus the translation established by one of the matches and the given flexion and rotation angles, completely determined the pose of the knee for the contact 4-tuple. In the notation of Fig. 4, given
Fig. 4. Varus angulation is determined by the contact 4-tuple locations.

contact locations \([(y_1, z_1), (y_2, z_2)]\), the numerical value of varus angulation is

\[ V = \tan^{-1}\left(\frac{z_2 - z_1}{y_2 - y_1}\right). \]  

In practice, because the surfaces (and associated normals) were represented as sets of discrete points, exact matches would not in general be found. Let \( \delta \) be a distance tolerance (in millimeters), and let \( \epsilon \) be an angular tolerance (in radians). Using the fact that the angle \( \theta \) between two unit vectors \( a \) and \( b \) satisfies the equation \( \cos(\theta) = a \cdot b \), Eqs. (3)–(6) could be restated for discrete measurements as

\[
\|\hat{C}_m - R(C_m) - \hat{d}\| \leq \delta, \tag{8}
\]

\[
\|\hat{C}_i - R(C_i) - \hat{d}\| \leq \delta, \tag{9}
\]

\[
\hat{n}_m \cdot (-R(n_m)) \geq \cos(\epsilon), \tag{10}
\]

\[
\hat{n}_1 \cdot (-R(n_1)) \geq \cos(\epsilon). \tag{11}
\]

A transformation that satisfied Eqs. (8)–(11) unified the coordinate systems, so that for a given contact 4-tuple there was a transformation from (mobile) femoral coordinates to (absolute) tibial coordinates.

2.3. Articular geometry

The accuracy of our model simulation depends on the representation of the articular geometry. Ideally, the articular geometry should be derived directly from the manufacturer’s mechanical designs. These were not available to us when we initiated the study, so we obtained the articular geometries from three-dimensional laser surface scans, which reverse-engineered the articular surfaces into point clouds with associated point normals. The sampling resolution of the laser scanner used was about 0.5 mm. This potentially introduced an a priori source of error: the location of the surface points, and the associated point normals, may deviate from what they might have been if they were derived from the original drawings. Moreover, the laser scanning process did not provide uniform sampling resolution: due to the high curvature of the femoral component, its surface was sampled in three passes. Fig. 5 depicts a region of femoral component that was sampled nonuniformly.

The consequence of coarse, nonuniform surface sampling was inaccurate estimation of the local surface normal, which required a rather large numerical tolerance when matching surface normals. Such error is propagated through the computations, so the computed ligament strain minimum did not necessarily occur at the true minimum location.

This source of inaccuracy was addressed in two ways. First, the point normals obtained directly from the laser scanning process were disregarded, and re-computed with the following procedures: the given point clouds were triangulated into surface-based meshes that were entirely composed of triangles. The point normals associated with each point in the point cloud were then re-computed as the average of face_normals of all triangles that included the point in question. Such a smoothing process resulted in better estimation of the point normals in the given point clouds.

The second method to improve the accuracy is to compute a contact patch, rather than computing an ideal contact point. The contact patch was the set of contact locations that resulted in ligament strain energy no more than a user-defined tolerance higher than the value computed at the energy-minimizing contact location. The instantaneous contact location was calculated as the centroid of the contact patch.

2.4. Ligament strain energy

Ligaments were modeled as sets of independent, straight filaments. Each filament had a neutral length, and stored no strain energy if the distance between the femoral and tibial
The computation of knee kinematics took place in two stages. *Off-line*, the set of all feasible contact 4-tuples was determined for a range of prosthesis poses. *On-line*, patient’s ligament insertion locations were measured and used to determine the instantaneous contact location that minimized ligament strain energy for each prosthesis pose; because there were only a few thousand contact 4-tuples per pose, the minimum strain energy could be determined by exhaustively examining the strain energy at each pose with little computational effort. Knee kinematics were then animated as a sequence of instantaneous contact locations by graphically computing the femoral/tibial poses, and the ligament states, for each contact state (smoother animation is achieved simply by linearly interpolating between poses).

The advantage of the separation of purely geometrical constraints from a patient’s biomechanics is two-fold. First, the off-line computation reduces the real time needed to determine the patient-specific kinematics. Second, various surgical strategies, including different component placements and ligament releases, could be simulated easily with this model.

### 3. Validation and results

To validate our knee model, we first created a standard knee that was symmetric about the central sagittal plane. The articular geometry used is size 3 of the Anatomic Modular Knee (AMK) of DePuy Inc. (Warsaw, IN, USA). Ligament data were derived from our previous two-dimensional study on a series of patients (Martelli et al., 1998), with ligament insertion locations adjusted so that the MCL was the mirror image of the LCL and so that the PCL was located midway between the collaterals.

In that previous study, the positions and orientations of the prosthetic components, and the positions of the ligament insertion sites, were measured with respect to the resection planes (taken at the time between verification of the trial implants and installation of the final implants, when the nurses were transferring the components to the surgeon). Five points were sampled from each of the three major resections and the points were fit to planes in the least-squares sense. (The points were the corners of each resection and a point substantially in the center of each resection.) The prosthesis was fit to the deduced resection planes by aligning the distal planes exactly and then fitting the line segments of intersection of the anterior and posterior planes of the data to the corresponding lines of the prosthetic model by linear averaging.

The neutral lengths of the ligaments were determined physically, by gently distracting the tibia until the relevant ligaments were determined to be at neutral length by the senior surgeon of the previous study (Dr. M. Marcacci). The PCL, MCL and LCL were represented by three, four and four independent filaments, respectively. A symmetrical knee was used for validation because the expected motion would occur only in the anterior–posterior direction, so any other motions would indicate a problem with the software.

The articular geometries were virtually implanted to a standard position, as recommended by the manufacturer’s description of surgical technique and later validated by a surgeon (JFR). The model was then tested with 1575
predicted that the PCL would be relaxed at low flexion. In near-extension poses, the knee was held in place by the collateral ligaments, which initially contributed the majority of the total ligament strain. The PCL would gradually elongate and become the predominant constraint of knee motion after approximately 65° of knee flexion, which is anatomically appropriate. At the same time, the collateral ligaments would gradually relax. At deep flexion the PCL would be stretched, causing the femoral component to move posteriorly. This prediction is consistent with reported results (Essinger et al., 1989), in which the PCL fibers were shown to be elongated during knee flexion to as much as 1.4 times the original length at 120° of flexion.

The range of postoperative flexion was limited by PCL strain. Based on previously validated criteria (Garg and Walker, 1990; Butler et al., 1986), a numerical value of 10% was chosen for the PCL strain that limited knee flexion to 120° in 5° increments. Flexion/extension angles ranged from 0° to 120° in 5° increments. Internal/external rotation angles ranged from −9° to +9° in 3° increments. Varus/valgus angle ranged from −2.0° to +2.0° in 0.5° increments. At each angulation, the contact state that minimized the total strain energy in the ligaments was noted, as were the individual ligament strains. For each flexion angle, the varus/valgus angle and internal/external rotation angle that jointly had the minimal total strain energy were selected as the orientation of passive motion, producing a total of 63 angulations of passive motion.

3.1. Ligament strain and joint laxity

The instantaneous contact location for each pose was determined from the strain energy stored in the posterior cruciate ligament and collateral ligaments. The contribution from each ligament varied through knee flexion. The ligament states of a knee with standard surgical placement of prosthetic components is plotted in Fig. 6, and a mid-flexion view of the components and the ligaments is shown in Fig. 7.

For the standard surgical placement, our model predicted that the PCL would be relaxed at low flexion. In near-extension poses, the knee was held in place by the collateral ligaments, which initially contributed the majority of the total ligament strain. The PCL would gradually elongate and become the predominant constraint of knee motion after approximately 65° of knee flexion, which is anatomically appropriate. At the same time, the collateral ligaments would gradually relax. At deep flexion the PCL would be stretched, causing the femoral component to move posteriorly. This prediction is consistent with reported results (Essinger et al., 1989), in which the PCL fibers were shown to be elongated during knee flexion to as much as 1.4 times the original length at 120° of flexion.

The range of postoperative flexion was limited by PCL strain. Based on previously validated criteria (Garg and Walker, 1990; Butler et al., 1986), a numerical value of 10% was chosen for the PCL strain that limited knee motion. For standard surgical placement of components, this corresponds to a range of postoperative motion from 0° to about 95° flexion.

3.2. Contact path and passive kinematics

The contact path on the tibial component is of considerable interest to surgeons and engineers because the wearing characteristics of the polyethylene bearing surface depend on how it is loaded over time. Fig. 8 depicts the trajectory of the contact path on the tibial bearing surface.

For the knee with standard component placements, the tibial contact was located at the anterior center of the tibial plateau at full flexion. Spinning motion was observed at low flexion angles (<30°). The tibial contact location traveled posteriorly as the flexion angle increased. In mid-flexion, the distance between successive contact locations was almost uniform, suggesting that a rolling motion occurred. In deep flexion, the tibial contact locations were located at the posterior end of tibial plateau. Overall, the posterior/anterior movement of the contact locations through flexion/extension angulation, respectively, corresponded to roll-back and roll-forward motions, which are also observed in normal postoperative gait motion. This pattern of gait motion suggested that polyethylene failure, of the type associated with spinning in place, should occur at the extreme anterior and posterior margins of the prosthesis.

3.3. Kinematics animation

Knee motion can be animated using standard computer graphics. Fig. 9 depicts the planner knee kinematics from full extension to deep flexion. It confirmed that the prosthesis moved in a spinning–rolling–spinning pattern. It also demonstrated an anatomically correct behavior known as ligament cross-over (Fig. 7(b)): at high flexion angles, the filaments in the medial collateral ligament of
3.4. Alternative surgical implantation strategies

There is evidence, from clinical follow-up studies, that the surgical placement of components can affect the postoperative range of motion achieved (Garg and Walker, 1990). To achieve better outcomes, surgeons may choose to implant prosthetic components at different positions or orientations than those suggested by the manufacturer. We simulated the effects of five alternative surgical placements. From the standard placement, the tibial component was displaced by 5 mm in each of the anterior and

Fig. 8. Passive kinematics of the standard knee. (a) The path of the contact centroid on the tibial component. (b) The relative location of the contact centroid on the femoral and tibial components describe the local kinematics; pure spinning and pure rolling are depicted as lines in the plot.

Fig. 9. Snapshots from the animation of passive knee kinematics.
posterior directions. From the neutral placement, the tibial component was also sloped 5° in the anterior and posterior directions. Lastly, from the posterior translation of 5 mm, the tibial component was sloped by 5° to further lower the posterior aspect of the bearing surface.

Fig. 10 depicts the ligament state, contact path and knee kinematics of a knee with a posteriorly displaced tibial component. The ligament strains were slightly increased at low flexion angles and slightly decreased in deep flexion when compared to the standard knee; the increased constraint at low flexion angles was mainly due to increased strain in the collateral ligaments. The contact locations at low flexion angles were slightly more concentrated at the anterior portion of the tibial plateau. The contact location moved posteriorly through mid-flexion, not much changed from the standard knee, and reached the most posterior periphery of the tibial bearing surface in deep flexion. It is clear that posterior displacement of the tibial component slightly increased spinning at low flexion angles and slightly decreased spinning in deep flexion.

Fig. 11 depicts the effects of displacing the tibial component anteriorly. The ligament strains were decreased at low flexion angles and much increased in deep flexion; there was less constraint at low flexion angles as the contributions of the collateral ligaments were decreased, and much greater constraint by the PCL in deep flexion.

The contact locations at low flexion angles were noticeably changed — they were dispersed, rather than being concentrated. A greater medial/lateral width was traversed in mid-flexion, suggesting that a different part of the femoral bearing surface was engaged with the tibial bearing surface. There was an increased concentration of contact at the posterior periphery in deep flexion. Anterior displacement of the tibial component also almost entirely eliminated low-flexion spinning and instead produced a rolling motion that continued through mid-flexion. The posterior spinning was initiated at a lower flexion angle than occurred for the standard knee.

Figs. 12 and 13 depict the effect of tilting the tibial component posteriorly and anteriorly, respectively. Posterior sloping of the tibia had little effect on the ligament states at early flexion angles. There was, however, a slight decrease in constraint through the latter part of mid-flexion and a definite decrease in constraint in deep flexion that was due principally to lower strain energy in the PCL. The anteriorly sloped tibial component produced little change in early flexion and mid-flexion, with a substantial increase in PCL constraint in deep flexion. Neither sloping had a discernible effect on ligament strains at low flexion angle, and the sloping had considerable (opposite) effects in deep flexion. The overall pattern of the contact paths of the sloped tibia was similar to those of the standard knee;
however, the contact paths for the posteriorly sloped tibia were translated posteriorly and those of the anteriorly tilted tibia were translated anteriorly. The overall kinematics of the knee with a posteriorly sloped tibia were the same as those of the standard knee: spinning in place at low flexion was followed by rolling in mid-flexion and ended in spinning. The range of the displacement over the tibial component was increased, suggesting that posterior sloping led to greater coverage of the tibial plateau. Anterior sloping of the tibial component had little discernible effect on passive knee kinematics.

Fig. 14 depicts the effects of a knee with a tibial component that is both posteriorly sloped and displaced. This can occur clinically, for example when the surgeon decides to use a different size of tibial component than that recommended by the manufacturer and also introduces a posterior slope. The simulation suggested that the ligament strains were increased in early flexion and substantially
Table 1

Comparison of kinematics resulting from different tibial-component placements

<table>
<thead>
<tr>
<th>Tibia placement</th>
<th>Kinematics</th>
<th>Low flexion</th>
<th>Mid flexion</th>
<th>Deep flexion</th>
<th>Range of motion</th>
</tr>
</thead>
<tbody>
<tr>
<td>Standard</td>
<td>Spin</td>
<td>Roll</td>
<td>Spin</td>
<td>0°–95°</td>
<td></td>
</tr>
<tr>
<td>Anteriorly displaced</td>
<td>Roll</td>
<td>Roll</td>
<td>Extended spin</td>
<td>Reduced</td>
<td></td>
</tr>
<tr>
<td>Posteriorly displaced</td>
<td>Extended spin</td>
<td>Roll</td>
<td>Reduced spin</td>
<td>Extended</td>
<td></td>
</tr>
<tr>
<td>Anteriorly rotated</td>
<td>Spin</td>
<td>Roll</td>
<td>Spin</td>
<td>Reduced</td>
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<tr>
<td>Posteriorly rotated</td>
<td>Spin</td>
<td>Roll</td>
<td>Extended spin</td>
<td>Extended</td>
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</tr>
<tr>
<td>Posteriorly rotated &amp; displaced</td>
<td>Spin</td>
<td>Roll</td>
<td>Spin</td>
<td>Greatly extended</td>
<td></td>
</tr>
</tbody>
</table>

decreased in deep flexion. This produced a wider range of motion, because the range of motion is primarily limited by PCL strain. The contact path displayed a slightly increased concentration of anterior contact, and a more widespread range of contact on the central and posterior bearing surfaces. Knee kinematics were noticeable changed: the range of spinning during early flexion was increased by about 10°, and the spinning in deep flexion was decreased by about 15°. The overall result is an increase of rolling motion through mid-flexion.

Table 1 summarizes the effects of alternative surgical placement of the tibial components. The knee with an anteriorly displaced tibial component had different knee kinematics than the standard knee. Predicted kinematics for low flexion angles was a rolling motion instead of a spinning motion. The range of motion was also reduced. This is because anterior displacement of the tibial component effectively increased the distance between the femoral and tibial PCL insertion sites, which led to a highly elongated PCL at high flexion. Similarly, the anteriorly sloped tibial component had reduced range motion, because anterior sloping increased the distance between ligament insertion sites. However, such effects were minimal at low flexion angles, for which the contact locations were principally in the anterior region of the tibial component.

Posterior displacement or sloping of the tibial component increased the range of motion, because such surgical placements of tibia decreased the distance between PCL insertion sites. Of particular interest, the knee with both a posteriorly displaced and a posteriorly sloped tibial component had greatly increased range motion, with the overall kinematics remaining as spin–roll–spin motion.

3.5. A surgical retrieval

Over 90% of total knee prostheses are still installed at 5 years after operation. In cases of aseptic loosening or failure of knee components, the principal suspects are improper alignment and improper balancing of the ligaments. One of us (JFR) is a practicing orthopedic surgeon who revises the knees of patients because of prosthesis failures. Recently, he revised the knee of a patient in whom an AMK prosthesis was implanted. The tibial component was retrieved, postoperatively decontaminated, and examined to determine the wear patterns.

Third-body wear particles, most likely particles of the bone cement used to hold the prosthesis in place, exaggerated the wear patterns. Our previous work (Cornwall et al., 1995) demonstrated the strong correlations between the kinematic conditions of contact and the resulting surface-degradation patterns on a tibial component. Fig. 15 shows the wear patterns on the tibial bearing surface. Burnishing and scratches are consistent with spinning motion. Cold flow is usually associated with creep under high contact stress, which can occur during spinning motion. Pitting is often associated with fatigue wear under high stress conditions while the contact region is moving, which can occur with rolling or sliding motions. We cannot definitively determine the surgical placement of the original prosthesis, nor can we draw statistical conclusions from a single retrieval, but the wear patterns are consistent with the general kinematics of the knee that our model predicts.

3.6. Applications

Due to the separation of off-line and on-line computations, knee kinematics from full extension to deep flexion can be calculated in just a few minutes for any ligament/prostheses pair. This allows several possible applications of our knee model.

Preoperative surgical planning can be improved if a patient-specific biomechanical model is available. Provided that a patient’s knee data can be obtained prior to surgery, the surgeon then has the ability to determine the most suitable prosthesis design, along with the proper surgical placement, for a patient. Effect of mal-sized femoral components, sloping of the tibial insert, can be visualized with the simulated knee kinematics.

Intraoperative surgical adjustment can also benefit from simulation results. Ligament release is often performed during surgery, and because of the modular design of most prostheses the surgeon often has the ability to select a tibial insert with different thickness. Effects of ligament release, and tibial insert thickness, can be visualized in real time by intraoperative simulation.

The model may also help in the process of prosthesis design. Some current knee prostheses are designed to have
non-congruent bearing surfaces. Effects of different design parameters, such as the flatness of the tibial insert and the radii of the femoral head, can be simulated using our model. Over time, a database of ligament data can also be built, and some statistical model of how a knee design parameter affects the knee motions can be developed as well.

4. Conclusions

We have described the development and construction of a computational model of the kinematics of condylar-type total-knee prostheses. The model required knowledge of the geometry of articular surfaces, the insertion locations and relative mechanical properties of the knee ligaments, and the implanted location of the prosthetic components. Based on the principle of ligament strain minimization, the computer program generated the motion of the center of contact of each condyle, the kinematics of the prosthetic joint, and the resulting ligament state from full extension to deep flexion.

The model was first validated on a prosthesis design for which the lateral profiles were the mirror images of the medial profiles. Using the articular surfaces of an AMK prosthesis and ligament data derived from previous work (Martelli et al., 1998), the model was used to simulate the spatial kinematics of a prosthetic knee. The model demonstrated that the pattern of knee motion was influenced by the design of the prosthesis and by the surgical implantation strategies of the prosthetic components. With respect to the AMK design, the simulation suggested that spinning and point-loading motions take place; these predictions are consistent with clinical observations and surgical retrieval of failed implants. The predicted motion is very similar to motion observed in vivo by Banks et al. (1997) under fluoroscopy. An additional observation we make on their work is that the motion they refer to as “paradoxical roll-forward” in early flexion may instead be spinning in place, and that in some cases the apparent roll forward may be a visual illusion (as can be seen in our animation).

The model provided a unique tool for evaluating the effects of different surgical implantation strategies. The model predicted that posterior displacement or sloping of the tibial component would usually lead to improved patterns of knee kinematics and increased postoperative ranges of motion. Conversely, anterior displacement or sloping of the tibia would have adverse effects, such as excessive elongation of the PCL and point loading in deep flexion. These findings are consistent with current knowledge of the in vivo behavior of this prosthesis design (Garg and Walker, 1990; Whiteside and Amador, 1988).

Our model is similar to the model of Essinger et al., in that both models compute the postoperative kinematics of the knee by minimizing the strain energy stored in the relevant ligaments. The main differences are that we developed our model for fast computations of passive kinematics, and so we (a) assumed that the bones and components were rigid, (b) normalize the mechanical properties of the ligaments, and (c) use a point-based exhaustive search for the global minimum of the total strain energy. Essinger et al. were interested in modeling active motion, so their model is more detailed (e.g., they model the quadriceps mechanism) but uses complex energy...

Fig. 15. A surgically retrieved AMK tibial component. The wear patterns were exaggerated by the presence of third-body wear particles. These wear patterns are consistent with spinning at low flexion angles, rolling in mid-flexion, and spinning or sliding in deep flexion.
calculations that are not suited to rapid calculation of the solution. The models generally agree in their predictions of knee motion.

Our model is less similar to the model of Blankevoort et al., because they solve force/moment equilibria to determine knee motion. Both models handle ligaments as groups of multiple fibers, and both models describe knee motion as a set of quasi-static instantaneous solutions. Our model does not have as many model parameters to tune, and thus may be more robust than their model. Because we model the surface of the components as piecewise-planar facets, our model does not have the difficulties in satisfying contact conditions of the protheses that they report.

There are shortcomings of this preliminary model and its validation. Although the model was validated by simulating the geometry of the knee of an actual patient, the lateral collateral ligaments were simply the mirror images of the medial collateral ligaments. This was useful in confirming that the computations were correct and that the overall motion was reasonable. The model does not account for important effects (such as the “screw-home” mechanism) that may occur from anatomically correct ligament models. The ligaments were modeled as composed of straight filaments; a better approach is to model the ligaments so that they follow the epicondylar contours. Further analyses of other patient-specific ligament models and surgical retrievals are also needed to establish that the predicted kinematics matches wear patterns produced in vivo.

It must also be recognized that there is a distinction between passive kinematics and active kinematics. Passive kinematics is technically the knee motion without regard to masses and external force and is functionally the knee motion examined by a surgeon when the patient is relaxed. Active kinematics is technically the knee motion that accounts for masses and external force and is functionally the knee motion activated by muscles and external loads. This study extends our previous 2D work on the relation between a computer model of passive kinematics and assessment of patients, and we cannot conclude that our model of passive kinematics accurately describes active kinematics. Except in laboratory settings, surgeons typically assess active kinematics only qualitatively but regularly identify passive kinematics semi-quantitatively (e.g., from assessment of laxity and range of motion) so the present study has focussed only on passive kinematics. Extension of our work to active kinematics/dynamics is potentially a valuable addition to the understanding of human motion.

Intraoperative surgical adjustment can also benefit from simulation results. Ligament release is often performed during surgery, and because of the modular design of most prostheses the surgeon often has the ability to select tibial insert with different thickness. Effects of ligament release, and tibial insert thickness, can be visualized in real time by intraoperative simulation.

Further validation of this model by in vitro and in vivo experiments is needed, particularly to consider the effects of asymmetric ligaments and asymmetric components. The model may prove useful not only to determine patient-specific implantation parameters but also to design new knee components that interact better with the biological structures of the human knee. Further work may include the use of three-dimensional medical images, particularly MRI, to preoperatively extract the anatomical data needed for a preoperative prediction of patient-specific postoperative knee motion.

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