A Technique for Accurate Kinematic Validation of a Subject Specific TKA Model

Willy Theodore1, 2, Joseph Little1, Edgar A Wakelin1*, Justin P. Roe3, Darryl D’lima4, Mark Taylor1, Brad Miles2

1Medical Device Research Institute, College of Science and Engineering, Flinders University, Adelaide, SA 5042, Australia
2360 Knee Systems Pty Ltd, Suite 3 Building 1, 20 Bridge Street Pymble NSW 2073, Australia.
3North Sydney Orthopaedic and Sports Medicine Centre, the Mater Hospital, NSW, Australia.
4Shiley Center for Orthopaedic Research and Education at Scripps Clinic, La Jolla, CA, United States.

Abstract

Background: High complexity computational models that require invasive data capture may provide high kinematic accuracy but have limited clinical applicability in total knee arthroplasty. Such models rely on validation procedures which do not compare model and experimental outputs with identical definitions. The objective of this study was to develop a validation method that allows cadaveric and model-predicted kinematics to be compared in the same frame of reference, and to apply the method to a simplified subject specific knee computational model (KCM).

Methods: Deep knee bend kinematics of eight cadaveric specimens were measured using a simulator based on the Oxford-knee-rig. A KCM was developed for each specimen to replicate the experimental boundary condition. Cadaveric experimental data were transformed to the computational model reference frame allowing comparison of kinematics with identical definitions, generating Virtualised Experimental Kinematics (VEK). Model-predicted kinematics were compared to these kinematics for validation.

Findings: The KCM developed in this study demonstrated good agreement with VEK. Tibiofemoral kinematic RMS differences were 0.5°SD0.17° coronally, 1.44°SD0.51° axially, and translational RMS differences of 1.22SD0.33 mm coronally, 0.53SD0.25 mm axially, 1.74SD0.5 mm sagittally. Using the validation method described, data collection errors such as unrealistic tibio-femoral penetration and patella-femoral separation were identified.

Interpretation: Validation of computational models using identical spatial and kinematic definitions is possible. A relatively simplified KCM was shown to achieve good kinematic agreement with experimental data, and compare favourably to previous studies while maintaining computational efficiency. Future validation studies need to place careful consideration on experimental procedure to prevent compound errors masking the accuracy of computational models.

Keywords: Total Knee Arthroplasty; Computational Modelling; Kinematics; Joint Mechanics; Model Validation

Introduction

Total Knee Arthroplasty (TKA) is a highly successful surgical intervention that relieves pain and improves the quality of life for osteoarthritis sufferers. Despite low revision rates, as many as 20% of recipients remain dissatisfied with long term pain after surgery [1-3]. Component alignment [4-6] as well as patient specific anatomical variation [7-9] has been shown to alter the kinematics and performance of the knee. Knee kinematics post-TKA therefore, is a product of complex interactions between the component design, component alignment, and patient specific anatomical characteristics [10-12]. Understanding these complex interactions pre- and intra-operatively can help surgeons plan surgery to achieve optimal outcomes.

Techniques such as ex-vivo force-controlled mechanical simulators [13-16], video fluoroscopy [17] and gait labs [18] have been used to study knee kinematics under simulated and real dynamic activities. These processes however, are expensive and inefficient to iterate experiments over a wide variety of variables (i.e. alignment and position of each component, variable anatomy etc.). Computational modelling however, is a scalable alternative [19] that allows the study of both patient and surgical factors and their interactions on knee kinematics.

Several knee computational model studies have been reported academically in literature, with various degrees of complexity [20-25]. Kuriyama et al. and others have generated knee simulations using generic anatomy in which bone models are scaled for patient specificity [26-28], while others have used 3D imaging when defining anatomy and subject specific ligaments inputs [21, 29, 30]. Generally, high complexity models require a greater number of inputs and computing power to simulate knee performance, leading to a trade-off between utility and complexity. Complexity of the ligament representation in computational models range from non-linear springs [26, 31-33] to
A recent study by Beidokhti et al. [34], found that although subject specific modelled ligaments improved contact accuracy, when joint kinematics are of most interest, spring ligament models incorporating literature-based parameters provide better computational efficiency with acceptable outcomes. Open source platforms such as Open Knee have the potential for wider clinical use, however to date, routine pre-surgical planning using patient specific simulations has not been achieved.

The validity of each of these models is critical for its utility and is usually performed against an ex-vivo cadaveric test that mimics model boundary conditions [37]. Previous validation studies have compared model kinematics against kinematics recorded in an experiment knee simulator [22, 30, 34, 38-41] and are usually described in the Grood and Sundan coordinate system [42]. Although these studies report experiment and model joint kinematics using the similar definitions, it is unclear whether the landmarks used to create the coordinate system were identical and reliable. For example, Marra et al. [23], developed a subject specific knee computational model in which the subject bones and soft tissue attachment sites were scaled from a generalized library. Two co-ordinate systems therefore, were defined from bone landmarks and skin based markers during validation, introducing error in both soft tissue force vectors and axis definitions. Baldwin et al. [22, 43], have reported validation results for finite element computational knee simulators in which a single computational frame of reference was defined according to digitized surface point data, and digitized anatomic references. TKA navigation studies however, have shown that digitization of anatomical landmarks in a surgical setting can be unreliable [44], indicating that kinematic comparison between two specimen may mask the real performance of a model. Kia et al. [21], developed a method in which a single reference frame was generated by digitizing and registering a radio-opaque artificial surface fixed to the femur and tibia in the CT scan. However, other anatomic landmarks defining soft tissue attachment sites were digitized during dissection, limiting the clinical applications of such a model.

In this study, we present a method to transform experimental outputs from a mechanical simulator to apatient specific reference frame originating from a computational model in which the only digitization points are from the implanted components. This ensures identical non-digitized anatomic landmark definitions when comparing model-predicted and experiment kinematics, eliminating a source of error identified in previous validation procedures. A secondary objective of this work was to use the described method to validate a CT-based subject-specific computational model that incorporates literature-based soft tissue properties. The model was developed to have sufficient complexity to achieve accuracy in kinematics when compared against experimental data while limiting invasive data capture to improve clinical utility. Cadaveric testing was conducted using a mechanical simulator that replicates the Oxford Knee Rig (OKR), performing a deep knee bend [15, 45].

**Method**

**Cadaveric Testing**

Ten fresh-frozen human cadaver knees underwent pre-operative CT scans of 1mm slice thickness and spacing (Aquilion, Toshiba). TKA was then performed on each of the cadaver knees with fixed-bearing, ultracongruent (with PCL sacrificed), TKA components (GMK, Medacta, Switzerland), after which a post-operative CT was obtained.

A surgical navigation system that utilised infrared optical tracking of LED arrays (Stryker Navigation 4.0 Research Version, Kalamazoo, Michigan, USA) was used to record kinematics throughout a deep knee bend. These arrays were attached to the femur, tibia and patella. The implant geometry surface (IGS) of each component were then digitized, allowing the position of the implants and bone to be mapped relative to the navigation arrays respectively. Figure 1 shows the overview of experiment steps in a flow chart.

Each cadaver was mounted in a dynamic, quadriceps-driven, closed-kinetic-chain knee simulator based on the OKR design, as previously described [46]. Stainless steel threaded rods were cemented into the femoral and tibial medullary canals, fixing the bones to the testing apparatus. The femoral rod was attached with a lateral offset of 5° from the hip joint to reproduce the average anatomical valgus in the femoral shaft [46]. The ankle joint was fixed rigidly to prevent translation, but the rotational degrees of freedom were left unconstrained. The hip joint was positioned directly vertical to the ankle joint and allowed to translate vertically, as well as rotate in flexion-extension and varus-valgus. A vertical load was applied through the hip joint to generate a peak knee-flexion moment of approximately 40Nm, similar to that reported for stair climbing after TKA [46, 47]. The attached navigation arrays were tracked using the navigation unit during the second half of the deep knee bend cycle, i.e during extension.

**Computational Model**

A knee computational model (KCM), with subject specific inputs derived from a CT scan only and designed to replicate the experimental mechanical simulator, was developed in rigid body modelling software, ADAMS (MSC, California, US), as described in [48]. Image processing and segmentation was performed on both pre-op and post-op CT’s taken for each cadaver using ScanIP (Simpleware, Exeter, UK). The native geometry of the femur, tibia and patella were generated and anatomic landmarks were recorded from the pre-op CT on all bones to define local and global axes, as well as soft tissue attachments, similar to those defined by Theodore et al. [48], in which anatomic morphology and CT pixel intensity were used to inform landmark positions. The pre-operative bones and implanted TKA CAD geometries were then registered to the post-operative CT to obtain the relative position of the prostheses to the bones using +CAD (Simpleware, Exeter, UK). The transformation matrix for each registered pre-operative bone was then used to re-position the pre-operative landmarks in the post-operative reference frame, allowing accurate post-operative landmarking unaffected by metal component flare in the CT. The transformed pre-operative bone geometries, transformed landmarks and registered components were exported from +CAD into the KCM to generate a patient specific computational model, an example of which is shown in Figure 2.

The model includes the lateral collateral ligament (LCL), medial collateral ligament (MCL), patella tendon, quadriceps tendon and posterior knee capsule. The LCL was modeled with a single fibre bundle and the MCL was modeled with anterior and posterior bundles. The posterior capsule and quadriceps tendon were generated with origin and insertion points dictated by relevant landmark positions. The posterior cruciate ligament (PCL) was not included in the model because the tibia insert implanted required resection of the PCL.

All ligaments were modelled as non-linear springs according to the equations shown in Equation 1, where \( f \) is the axial force sustained...

Figure 1: Overview of experiment procedure.

Figure 2 (Left): The patient specific anatomy and implant position were determined from CT imaging. (Right) A schematic of KCM undergoing a deep knee bend. A vertical load was applied to the hip joint and the hip joint was positioned directly vertical to the ankle joint. The ankle joint was fixed rigidly in translation but has 3 degrees of rotational freedom.

by the ligament, $k$ is the stiffness parameter, $\varepsilon$ is the instantaneous ligament strain and $2\varepsilon_l$ is the threshold strain which dictates the change from the non-linear to linear regions [49]. The unstrained ligament lengths (free length) were calculated using a reference strain method by defining the ligament strain when the knee is at full extension according to Equation 2, where $\varepsilon_r$ is the ligament strain at full extension, $l_r$ is the length of the ligament at full extension and $l_0$ is the ligament free length. The free length is then used to calculate $\varepsilon$ in Equation 1. The reference strain and ligament stiffness values were obtained from previous studies [50-52] and are shown in Table 1.

The quadriceps tendon inserts in the midline of the superior patella and originates from the greater trochanter. The patellar tendon originates at the inferior apex of the patella and inserts at the medial third of the tibial tubercle. The patellar and quadriceps tendons force-length relationships were based on literature [53]. Contact was modelled between the: tibial insert and femoral component; femoral and patellar component; quadriceps tendon and femoral component; and between patellar tendon and tibial insert. All components were modelled as rigid bodies with kinematic and compliant constraints, using a penalty-based contact between components.

Equation 1: Mathematical description of ligament behaviour. Where $f$ is the ligament force, $\varepsilon$ is ligament strain, $\varepsilon_l$ the threshold strain at which the stress/strain behaviour of the ligament changes, and $k$ is the spring constant.

$$f = \frac{1}{4} k \frac{\varepsilon^2}{\varepsilon_l^{\frac{1}{2}}} \quad 0 \leq \varepsilon \leq 2 \varepsilon_l$$
$$f = k (\varepsilon - \varepsilon_l), \quad \varepsilon > 2 \varepsilon_l$$
$$f = 0, \quad \varepsilon < 0$$

Equation 2: Method of calculating reference ligament strain.

$$\varepsilon_r = \frac{l_r - l_0}{l_0}$$

Table 1: Details of ligament properties used in simulation [50-52].

<table>
<thead>
<tr>
<th>Ligament/tendon</th>
<th>Stiffness (N)</th>
<th>Reference strain at full extension</th>
</tr>
</thead>
<tbody>
<tr>
<td>Anterior MCL</td>
<td>2750</td>
<td>-0.06</td>
</tr>
<tr>
<td>Posterior MCL</td>
<td>2750</td>
<td>0.03</td>
</tr>
<tr>
<td>LCL</td>
<td>2000</td>
<td>0.05</td>
</tr>
</tbody>
</table>

The model was initialized in extension with a vertical load applied at the hip centre equal to that applied experimentally. During initialization of the model, all components were allowed to settle into contact, and ligaments were pre-strained. A deep knee bend was then simulated at the hip centre equal to that applied experimentally. During knee bend simulations required less than 3 minutes computational time to complete on a standard performance work station (i7- 4.00 GHz, 32GB RAM). Comparison of the model predicted subject specific tibiofemoral and patellofemoral kinematics with the VEK for a representative specimen (specimen 6) is shown in Figure 4. In general, the KCM was able to capture the trend and magnitude of translation and rotation across all measures.

Kinematics comparison and validation protocol

Figure 3 describes the steps taken to compare experimental and model-predicted kinematics. TKA CAD geometries were registered to best match the experimentally digitized IGS, allowing the digitized implant surface to be modelled as a whole component. The CAD geometries were registered to the implanted component position as observed in the post-operative CT during KCM generation. Experimental data were then transformed to the post-op CT frame of reference maintaining the relative position of the experimental femur and Patella CAD geometry to the tibia CAD geometry by normalizing the position of the tibia tray to that observed in the post-op CT.

Previously transformed pre-op CT anatomical landmarks were then duplicated and re-transformed accordingly to the respective positions of the experimental components. This gives 2 sets of registered components and landmarks for each specimen. The first set forms the subject specific model with anatomical landmarks from the pre-op CT and component positions as measured from the post-op CT. The second set is the virtualized experimental data extracted from the navigation system transformed to the post-op CT reference frame. Therefore a single set of anatomical landmarks, first generated from the pre-op CT only, define identical axes in the same relative position against all components in both the model and experimental data.

The positional data recorded by the navigation system during testing describes the relative kinematics of the tibia and patella against the femur. This data was applied to the digitized landmarks and components to recreate the motion recorded experimentally. The kinematics were then computed using the transformed anatomical landmarks using the Grood and Suntay definition, thus generating a set of Virtualised Experimental Kinematics (VEK) with identical definitions to the KCM.

To compare the overall difference between the VEK and KCM predicted kinematics, the root-mean-square (RMS) difference of each specimen was calculated between flexion angles of full extension and 90°. The average absolute difference at 10° intervals was also calculated, giving a measure of variability throughout regions of the flexion cycle.

Results

Experimental data from specimen 1 and 5 were incomplete and excluded from analysis, leaving 8 valid trials for analysis. All deep knee bend simulations required less than 3 minutes computational time to complete on a standard performance work station (i7- 4.00 GHz, 32GB RAM). Comparison of the model predicted subject specific tibiofemoral and patellofemoral kinematics with the VEK for a representative specimen (specimen 6) is shown in Figure 4. In general, the KCM was able to capture the trend and magnitude of translation and rotation across all measures.

The tibiofemoral and patellofemoral RMS differences between the KCM and VEK for each specimen between full extension and 90° flexion are presented in Table 2 and Table 3 respectively. RMS differences for all tibiofemoral kinematic measures show the KCM has good agreement with VEK, with a maximum RMS difference reported for tibiofemoral internal-external (IE) rotation of 2.11°. Tibiofemoral varus-valgus (VV) rotational kinematics show the lowest difference with an average RMS of 0.50°, followed by IE rotation with an average RMS of 1.44°. Component translational RMS differences show superior-inferior (SI) shift is reported most accurately with an RMS of 0.53 mm, followed by medial-lateral (ML) and anterior-posterior (AP) shift with an RMS of 1.22 mm and 1.74 mm respectively. The average tibiofemoral RMS difference in flexion-extension (FE) is 7.50°, which can be attributed to the large magnitude of the flexion values and gives similar results to the other axes when considered as a relative difference. Patellofemoral ML translation RMS was the lowest with a value of 0.87°, followed by AP shift with an RMS of 1.73 mm. IE rotation reports an RMS difference of 3.41°. Patellofemoral FE and SI RMS were similar to tibiofemoral flexion RMS with a difference of 6.62° and 5.59 mm respectively.
Figure 3: Generation of VEK and computational model kinematics comparison. Implanted TKA CAD geometries registered to both post-op CT and implant geometry surface (IGS) reference frame. Experimental data were transformed to the computational model reference frame using the transformation matrix sequence of the registered CAD geometries. Experimental data were firstly transformed to the CAD geometry reference frame using the inverse of the component registration transformation, then transformed to knee computational model using the transformation matrix from CAD geometry to post-op CT reference frame. Transformed experimental tibia CAD geometry was then normalized against the post-op CT tibia tray. Relative positional experimental data were then applied to femur and patella against the normalized tibia.
Figure 4: Comparison of tibiofemoral VEK and model-predicted kinematics for single representative specimen (specimen 6).

Table 2: Tibiofemoral (TF) KCM kinematics RMS difference compared to the VEK.

<table>
<thead>
<tr>
<th>Specimen</th>
<th>Tibiofemoral kinematics RMS</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>TF_VV</td>
</tr>
<tr>
<td>2</td>
<td>0.47</td>
</tr>
<tr>
<td>3</td>
<td>0.72</td>
</tr>
<tr>
<td>4</td>
<td>0.46</td>
</tr>
<tr>
<td>6</td>
<td>0.62</td>
</tr>
<tr>
<td>7</td>
<td>0.24</td>
</tr>
<tr>
<td>8</td>
<td>0.38</td>
</tr>
<tr>
<td>9</td>
<td>0.43</td>
</tr>
<tr>
<td>10</td>
<td>0.69</td>
</tr>
<tr>
<td>Average</td>
<td>0.50</td>
</tr>
<tr>
<td>Standard deviation</td>
<td>0.17</td>
</tr>
<tr>
<td>Max RMS</td>
<td>0.72</td>
</tr>
</tbody>
</table>
The average absolute difference between the KCM results and VEK between 0° - 90° flexion for tibiofemoral kinematics are less than 2 mm or 2° for all clinically relevant kinematic measures, shown as the bold lines in Figure 5A. Tibiofemoral VV are the most accurately predicted kinematics with an average absolute difference of less than 1°. IE rotation and AP shift have greater differences, reaching 1.33° and 1.97 mm respectively. The patellofemoral average absolute difference in kinematics from 0° - 90° flexion is shown in Figure 5B. ML shift and IE rotation are best modelled by the KCM with an average absolute difference of less than 1 mm and 4° respectively at all flexion angles. FE rotation and SI shift exhibit low absolute differences at low flexion angles (of 3.61° and 2.46 mm respectively). Differences in these measures increase from 50° flexion to a maximum of 8.34° and 7.87 mm respectively.

The average absolute difference between the KCM results and VEK between 0° - 90° flexion for tibiofemoral kinematics are less than 2 mm or 2° for all clinically relevant kinematic measures, shown as the bold lines in Figure 5A. Tibiofemoral VV are the most accurately predicted kinematics with an average absolute difference of less than 1°. IE rotation and AP shift have greater differences, reaching 1.33° and 1.97 mm respectively. The patellofemoral average absolute difference in kinematics from 0° - 90° flexion is shown in Figure 5B. ML shift and IE rotation are best modelled by the KCM with an average absolute difference of less than 1 mm and 4° respectively at all flexion angles. FE rotation and SI shift exhibit low absolute differences at low flexion angles (of 3.61° and 2.46 mm respectively). Differences in these measures increase from 50° flexion to a maximum of 8.34° and 7.87 mm respectively.

**Table 3:** Patellofemoral (PF) KCM kinematics RMS difference compared to the VEK.

<table>
<thead>
<tr>
<th>Specimen</th>
<th>Patellofemoral Kinematics RMS</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>PF_FE</td>
</tr>
<tr>
<td>2</td>
<td>12.82</td>
</tr>
<tr>
<td>3</td>
<td>4.48</td>
</tr>
<tr>
<td>4</td>
<td>7.66</td>
</tr>
<tr>
<td>6</td>
<td>4.30</td>
</tr>
<tr>
<td>7</td>
<td>2.04</td>
</tr>
<tr>
<td>8</td>
<td>8.12</td>
</tr>
<tr>
<td>9</td>
<td>4.93</td>
</tr>
<tr>
<td>10</td>
<td>8.58</td>
</tr>
<tr>
<td>Average</td>
<td>6.62</td>
</tr>
<tr>
<td>Standard deviation</td>
<td>3.37</td>
</tr>
<tr>
<td>Max RMS</td>
<td>12.82</td>
</tr>
</tbody>
</table>

**Figure 5:** Bold lines show the average absolute difference between the KCM kinematics and VEK from 0 and 90° flexion. Shaded region represents 1 standard deviation. Varus-valgus = green, Internal-external rotation = blue, Anterior-posterior shift = red.

**Recent Adv Arthroplast, 2019 Volume 3(1): 100 - 103**
Knee kinematics post-TKA are a product of complex interactions between the component design, component alignment, and patient specific anatomic characteristics [10-12]. Computational modelling is a scalable technology to study the complex interactions of these factors and can help surgeons better plan surgery to achieve optimal outcomes. However, such models need to balance clinical practicality with the complexity required to distinguish individual subjects. The KCM in this study was derived from subject-specific CT scans only, and literature-based values were used for the soft tissue representation. Good agreement was found in trend and magnitude between model-predicted and experimental kinematics in all 8 specimens using the method described. The kinematic differences observed in this study were comparable to previous studies [22, 31, 41, 56], particularly for tibiofemoral (TF) kinematics. Patellofemoral (PF) kinematics RMS difference was slightly higher in our model than previously reported by Baldwin et al. [22] which could be attributed to the simplification of the extensor mechanism in the model. The extensor mechanism was modelled with single point insertions of the quadriceps tendon and patella tendon, without retinaculum tissue or other medial and lateral structures. These simplifications appear to not affect predictions in ML shift and IE rotation (given the relatively lower RMS differences) but may explain some of the differences in PF flexion. The simplification however, was justified for computational efficiency, given that TF and PF ML shift and IE rotation are clinically more relevant. The KCM in this study completed a single deep knee bend within a few minutes, an efficiency that was not seen in more complex models [21, 22, 30, 34, 57].

Figure 6: Example images of IGS data (dots) with registered component geometries to the Femur (A), tibia tray (B), patella button (C) and whole knee (D). Penetration of approximately 1 mm between the femoral component and tibial insert can be seen in (E) in VEK during flexion.
Although a best fit algorithm was used to register the CAD geometries to the IGS points, some error remained, as seen in figure 6. The method described attempts to minimise the error in comparing kinematics, however as identified, inherent experimental or human error is still present in the data. All mechanical simulators have limitations [58], and when coupled with experimental procedure setup, the cumulative probability of unintentional errors increases. Apart from modelling and kinematic definition errors, validation results may also be affected by experimental data capture error. The infrared cameras used in the navigation system have intrinsic measurement error. The navigation arrays were assumed to be rigidly fixed to the bone, but could have unintentionally moved. Human error in digitizing, such as probes capturing points when not contacting the articular surface, combined with movement of the navigation arrays can affect the registration of the implant CAD geometries. Each of these sources of error will propagate throughout each calculation when generating the VEK. For example, the registered and transformed experimental patella button was observed at times to not always contact the femoral component during flexion. Similarly, as shown in Figure 6 the articular surface of the femoral component was at times observed to penetrate the tibial insert. In each case however, the registration error between the CAD geometries and the digitized surfaces was less than the error in separation or penetration of articular contact, indicating such errors arose from the data capture, rather than experimental post-processing.

There are a number of limitations to this study. Firstly, kinematic errors exist in cadaveric testing. The model-predicted kinematics may have been compared to experimental data with sequential data collection and registration errors. Error propagation such as this prevents differences between the KCM and VEK from being solely attributed to modelling errors. A more controlled experimental set-up that focuses on accurate digitization and better source of error quantification will help differentiate modelling error from experimental error and will be the subject of future work.

The KCM utilised a simplified model to represent soft tissue properties. Previously published validated computational knee models have conducted experimental studies to calibrate subject specific ligament properties. Baldwin et al. [22, 57], Ewing et al.,[29], and Mootanah et al. [30] performed passive laxity testing on cadaver specimens to determine characteristic VV and IE resistance. Inclusion of subject specific ligament characteristics may improve model accuracy, however, capturing patient specific ligament data is invasive and cannot be obtained in a clinical setting – preventing any such model from clinical use.

Model accuracy obtained from these studies therefore, will not necessarily translate to subjects in which invasive soft tissue data is not available. Beidokhti et al. has suggested that although subject specific modelled ligaments improved contact accuracy, when joint kinematics are of most interest, multi spring elements like those modelled here are the most efficient in terms of combining modelling accuracy with computational efficiency [34]. The KCM here was able to distinguish the most efficient in terms of combining modelling accuracy with computational efficiency [34]. The model-predicted kinematics improved contact accuracy, when joint kinematics necessarily translate to subjects in which invasive soft tissue data is not available.

Although a best fit algorithm was used to register the CAD geometries to the IGS points, some error remained, as seen in figure 6. The method described attempts to minimise the error in comparing kinematics, however as identified, inherent experimental or human error is still present in the data. All mechanical simulators have limitations [58], and when coupled with experimental procedure setup, the cumulative probability of unintentional errors increases. Apart from modelling and kinematic definition errors, validation results may also be affected by experimental data capture error. The infrared cameras used in the navigation system have intrinsic measurement error. The navigation arrays were assumed to be rigidly fixed to the bone, but could have unintentionally moved. Human error in digitizing, such as probes capturing points when not contacting the articular surface, combined with movement of the navigation arrays can affect the registration of the implant CAD geometries. Each of these sources of error will propagate throughout each calculation when generating the VEK. For example, the registered and transformed experimental patella button was observed at times to not always contact the femoral component during flexion. Similarly, as shown in Figure 6 the articular surface of the femoral component was at times observed to penetrate the tibial insert. In each case however, the registration error between the CAD geometries and the digitized surfaces was less than the error in separation or penetration of articular contact, indicating such errors arose from the data capture, rather than experimental post-processing.

There are a number of limitations to this study. Firstly, kinematic errors exist in cadaveric testing. The model-predicted kinematics may have been compared to experimental data with sequential data collection and registration errors. Error propagation such as this prevents differences between the KCM and VEK from being solely attributed to modelling errors. A more controlled experimental set-up that focuses on accurate digitization and better source of error quantification will help differentiate modelling error from experimental error and will be the subject of future work.

The KCM utilised a simplified model to represent soft tissue properties. Previously published validated computational knee models have conducted experimental studies to calibrate subject specific ligament properties. Baldwin et al. [22, 57], Ewing et al.,[29], and Mootanah et al. [30] performed passive laxity testing on cadaver specimens to determine characteristic VV and IE resistance. Inclusion of subject specific ligament characteristics may improve model accuracy, however, capturing patient specific ligament data is invasive and cannot be obtained in a clinical setting – preventing any such model from clinical use.

Model accuracy obtained from these studies therefore, will not necessarily translate to subjects in which invasive soft tissue data is not available. Beidokhti et al. has suggested that although subject specific modelled ligaments improved contact accuracy, when joint kinematics are of most interest, multi spring elements like those modelled here are the most efficient in terms of combining modelling accuracy with computational efficiency [34]. The KCM here was able to distinguish the most efficient in terms of combining modelling accuracy with computational efficiency. Further developments of this model and evaluation against supporting clinical data may lead to utilisation of computational modelling as part of computer assisted surgical planning for TKA.

Conclusions

In this study, a method that allows experimental and model-predicted kinematics to be compared with identical landmark and kinematic definitions was presented. Due to experimental error propagation, model validation studies need to place careful consideration on experimental procedure to prevent compound errors from masking the accuracy of computational models. Using the method described, a relatively simplified knee computational model was able to achieve good kinematic agreement while maintaining computational efficiency. Further developments of this model and evaluation against supporting clinical data may lead to utilisation of computational modelling as part of computer assisted surgical planning for TKA.

Acknowledgements

The authors thank Scripps Clinic for their support in the cadaveric testing and Medacta for providing implants and CAD geometries of the implant system. Also, the authors thank Brian Cheung and Scott Bergeon for their contribution to the model development.

References
