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Variability in static alignment and kinematics for kinematically aligned TKA

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A R T I C L E I N F O

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A B S T R A C T

Background: Total knee arthroplasty (TKA) significantly improves pain and restores a considerable degree of function. However, improvements are needed to increase patient satisfaction and restore kinematics to allow more physically demanding activities that active patients consider important. The aim of our study was to compare the alignment and motion of kinematically and mechanically aligned TKAs.

Methods: A patient specific musculoskeletal computer simulation was used to compare the tibio-femoral and patello-femoral kinematics between mechanically aligned and kinematically aligned TKA in 20 patients.

Results: When kinematically aligned, femoral components on average resulted in more valgus alignment to the mechanical axis and internally rotated to surgical transepicondylar axis whereas tibia component on average resulted in more varus alignment to the mechanical axis and internally rotated to tibial AP rotational axis. With kinematic alignment, tibio-femoral motion displayed greater tibial external rotation and lateral femoral flexion facet centre (FFC) translation with knee flexion than mechanical aligned TKA. At the patellofemoral joint, patella lateral shift of kinematically aligned TKA plateaued after 20 to 30° flexion while in mechanically aligned TKA it decreased continuously through the whole range of motion.

Conclusions: Kinematic alignment resulted in greater variation than mechanical alignment for all tibio-femoral and patello-femoral motion. Kinematic alignment places TKA components patient specific alignment which depends on the preoperative state of the knee resulting in greater variation in kinematics. The use of computational models has the potential to predict which alignment based on native alignment, kinematic or mechanical, could improve knee function for patient's undergoing TKA.

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1. Introduction

Total knee arthroplasty (TKA) is an established procedure for improving pain and restoring a significant degree of function, especially for low-demand activities of daily living. However, an understanding of optimal alignment and patient specific kinematics is needed to restore knee motion closer to normal, allowing performance of physically demanding activities that more active patients consider important [1–3].

The philosophy of mechanical alignment of the implant after TKA has traditionally been done to preserve longevity of the implant and enhance post-operative knee function [4–6]. However, studies have shown that although a mechanically aligned TKA improves the patient's function, 20% to 25% of patients remain dissatisfied [7,8]. In addition, recent data has challenged the importance of post-operative mechanical alignment in TKA. Paratte et al. [9], in a study reviewing 398 TKAs, demonstrated no improvement in the 15 year implant survival rate in patients within and outside of a post-operative mechanical alignment $0° \pm 3°$ (standard deviation).

Recently, kinematic alignment has been proposed by Howell et al. [10–14] as an alternative to restore normal knee motion and function. Kinematic alignment references the femoral transcylindrical axis, believed to be the flexion extension axis of the knee. The aim is to align the angle and level of the distal joint line of the femoral component, posterior joint line of the femoral component, and joint line of the tibial component to those of the normal knee [11].

Kinematically aligned TKA has been performed since 2006 however unanswered issues continue regarding patient outcomes, survivorship, surgical technique and use of specialised surgical guides [15–18]. A randomized controlled study demonstrated kinematically aligned TKA resulted in better pain relief, post-operative function and range of motion than mechanically aligned TKA in 88 patients (88 knees) [16]. Other studies emphasized higher function as assessed using the Oxford Knee Score and WOMAC™ score on 198 patients (214 knees) [17]; on 101 patients (101 knees) with kinematic alignment [18]. However, one small series emphasized the potential for malalignment using the OtisKnee system, which places implants at higher risk of early failure [15].

The optimal targets for alignment in TKA remain unclear, and indeed a single philosophy may not be applicable to an optimal outcome in all patients. Computer simulations are powerful tools that can provide insight into how different alignments influence post-operative outcomes for TKA patients. It allows control of component alignment for the same subject in ways not possible with in-vivo studies. With imaging data, computer simulations are also able to include patient variations into the analysis [19–22]. Previous studies with computational models have shown comparable kinematic and forces to those measured experimentally or with in-vivo fluoroscopy [23–26].

Ishikawa et al. [27] were able to analyse kinematic alignment for TKA using a computational knee simulation. Their study suggests that kinematically aligned TKA produces near-normal knee kinematics and may provide better clinical results than mechanically aligned TKA. However, only a single model was used in the study and the kinematic alignment for that single model was defined with the clinical average and therefore its conclusions were limited.

The aim of our study was to compare the alignment and motion of kinematically and mechanically aligned TKAs with a computational knee simulation using pre-operative Computer tomography (CT) scans from a series of 20 patients undergoing TKA. Computer simulation of both kinematic and mechanical alignments was performed for each subject. Measures of tibio-femoral translation, tibio-femoral rotation, patellar tilt and patellar shift were taken and compared between kinematic and mechanically aligned knees.

2. Materials and methods

2.1. Simulation set-up

A validated musculoskeletal computational simulation was used to evaluate the kinematic behaviour of kinematically and mechanically aligned TKA in a series of 20 subjects selected from 'The Joint Dynamics Registry' which includes

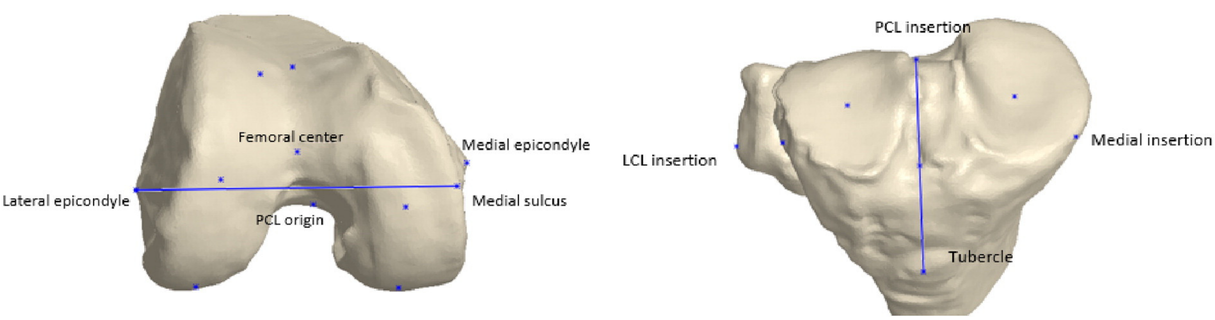

Figure 1. Schematic of landmarks and attachment points. Line connecting lateral epicondyle and medial sulcus defines the surgical transepicondylar (TEA) axis of the femur. Line connecting PCL insertion and tubercle defines the tibia anterior–posterior (AP) axis which then projected onto a plane perpendicular to the mechanical axis to be used as AP rotational axis as defined by Insall [29].

pre-operative CT scans of TKA patients (Bellberry Human Research Ethics Committee, approval number 2012-03-710). The simulation was developed using ADAMS MSC, California, a dynamic, quadriceps-driven, closed-kinetic-chain knee simulator based on the Oxford Knee Rig (OKR) [28]. Experimental validation results of the simulation model are provided in the Supplementary material.

Each model was assembled from CT scan segmentations of patient geometry using ScanIP segmentation software (Simpleware, Exeter, UK). CT scans were taken from degenerative joint diseased knees at a maximum of six weeks before scheduled TKA surgery. The population group had a mean age of 69.8 $\pm$ 7.3 years. Five of the patients were male and 15 were female. Of the simulated knees, eight were left knees and 12 were right. CT scans were taken at 1.25 mm slice thickness, with the other axial thicknesses varying but all less than 1.25 mm.

Landmarks were defined in order to assemble a patient specific model of relevant axes, ligament and tendon attachment sites associated with the reconstructed 3D patient geometry as shown in Figure 1.

The model includes the lateral collateral ligament (LCL), medial collateral ligament (MCL), posterior cruciate ligament (PCL), patella tendon, quadriceps tendon and posterior knee capsule. The LCL was considered to be a single fibre bundle and the MCL was considered to consist of anterior and posterior bundles. Likewise the PCL was modelled as an anterior and posterior bundle and was differentiated into anterior and posterior bundles by translation determined from experimental validation.

The femoral attachment points for the LCL and MCL were defined as the epicondylar prominences. The fibular LCL attachment was defined as attaching to the lateral–proximal centre of the fibular head. The tibial attachment points of the MCL bundles were modelled as attaching at the superior–inferior level of the peak medial prominence of the medial edge of the tibia distal to the plateau, with anterior–posterior position at the peak medial projection. The PCL's attachment on the femur was modelled as residing midway distally down the posterior intercondylar fossa when viewed from a posterior perspective, with the centre of attachment of the band placed one third of the width of the intercondylar fossa from the lateral edge of the medial condyle. Its tibial attachment was defined as the centre of the posterior intercondylar fossa.

The mechanical axis of the femur was defined as the line between the centre of the intercondylar notch to the centre of the femoral head, while the tibial mechanical axis was defined as the midpoint of the medial and lateral malleoli at the ankle to the midpoint of a line joining PCL insertion point and medial third of the tibial tubercle. The PCL insertion point and medial third of the tibial tubercle were then projected onto a plane perpendicular to the mechanical axis in order to define the tibial anterior–posterior (AP) rotational axis, as defined by Insall [28]. The surgical transepicondylar axis (the neutral femoral rotational axis) was defined by the lateral epicondylar point and the sulcus of the medial epicondyle.

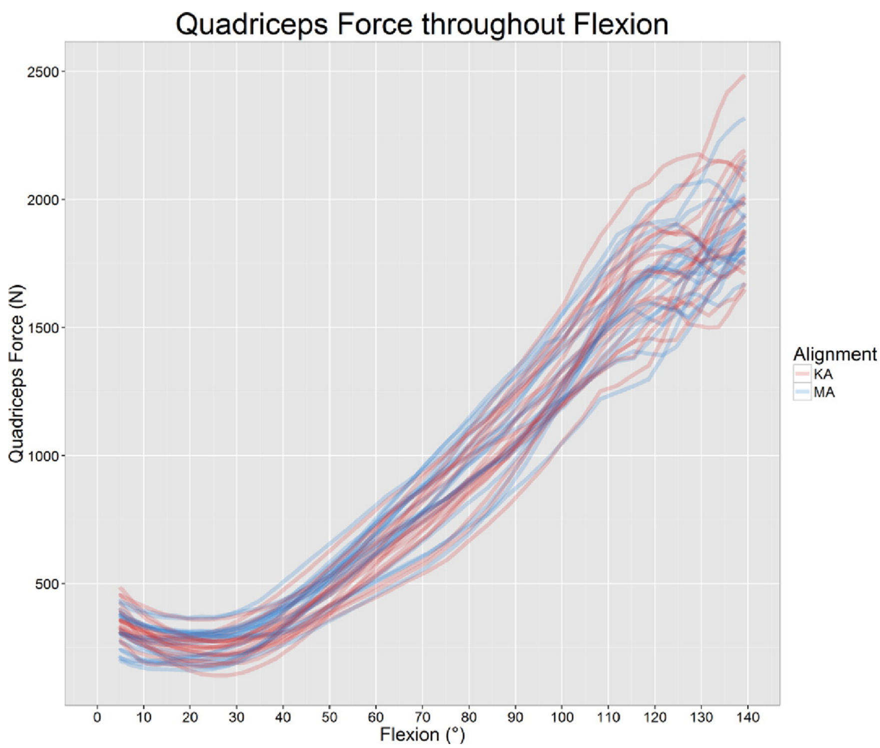

Figure 2. Quadriceps force throughout flexion for mechanical and kinematic alignments.

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Ligaments were modelled as point to point non-linear springs, shown in Eq. (1) [21].

$$
\begin{array}{ll}
f = \frac{1}{4}k\frac{\varepsilon^2}{\varepsilon_l}, & 0 \le \varepsilon \le 2\varepsilon_l \\
f = k(\varepsilon - \varepsilon_l), & \varepsilon > 2\varepsilon_l \\
f = 0, & \varepsilon < 0
\end{array}
\tag{1}
$$

where f is the axial force sustained by the ligament, k is a stiffness parameter, ε is the strain and $2\varepsilon_l$ is the threshold strain which indicates the change from the toe to the linear regions. The threshold strain used is adapted from literature [30]. The stiffness coefficients of the PCL, LCL and MCL were initially adapted from previous studies [21,30–32]. Ligament stiffness's were then adjusted based on experimental validation performed with a cadaver study. Initial pre-strain in each ligament in extension was assumed to match values reported previously in literature [30]. The patella tendon and quadriceps tendon were modelled as wrap-able segmented links with femoral component contact to allow for wrapping about the anterior femoral component in flexion.

2.2. The simulation

The simulation model simulated a closed-kinetic-chain knee extension based on the OKR. All components were modelled as rigid bodies with kinematic and compliant constraints, using a penalty based contact between components. The model initialised in extension and then the ankle joint was held rigid, which had three degrees of rotational freedom but was constrained in translation. The hip joint was positioned above the ankle joint and was allowed freedom in flexion–extension and varus–valgus, with the vertical motion guided by the axis drawn from the ankle–joint to the hip joint.

In the flexion cycle of the simulation, a negligible force was applied through the extensor mechanism to model soft tissue tension. Following the flexion cycle, the extensor mechanism was activated, using a force applied through the quadriceps tendon to drive the knee back into extension. A PID (proportional–integral–derivative) controller was used to generate the reactive quadriceps force required to achieve extension [26,33,34], as seen in Figure 2. The simulation runs through the flexion and extension cycle over a 10 s period, simulated using a dynamic multibody solver.

2.3. Mechanical and kinematic component placement

A fixed bearing, cruciate-retaining, symmetrical femoral and tibial condyle multi-radius implant design (Apex CR; OMNIlife science, East Taunton, MA, USA) was used to model both kinematic and mechanical TKAs for each of the 20 patients (Figures 3–5).

The mechanically aligned femoral components were aligned in the coronal plane perpendicular to the mechanical axis of the femur and rotated to be parallel with the projection of the surgical transepicondylar axis. Translationally, the femoral components were placed such that the most distal condyle of the native femur was level with the most distal point on the condyle of the implant, and likewise for posterior placement [35].

Femoral component flexion and size were then set by incrementally flexing the component until the anterior flange was flush to the anterior surface of the femur. A maximum of five degrees flexion was used as an upper limit before an upsized component was selected. Medial–lateral positioning was performed to result in equal amount of exposed bone on the medial and lateral sides.

The tibial component was placed perpendicular to the mechanical axis for all 20 mechanically aligned simulations and rotated to match tibial AP rotational axis defined above. The component was placed proximally to match the resection level of the thinnest tibia insert and had its medial–lateral and antero-posterior position chosen to maximize coverage subject to those orientations. Posterior slope for all tibial components was set at three degrees from a line perpendicular to tibia mechanical axis.

For the kinematically aligned knees, the femoral component was positioned such that the distal and posterior condyles of the femoral component match the joint line of the native femur. The component was then flexed and upsized as needed to avoid femoral component notching. For the tibial component, rotation was defined by a best fit ellipse drawn on the lateral plateau of the tibia in order to replicate the intra-operative technique described by Howell et al. [36] (Figure 6). Posterior slope of the tibial component was set at three degrees less than the posterior slope of the native medial condyle. Coronal plane alignment was set level to tibia joint line and proximalised to match the resection level of the thinnest tibia insert. Medial lateral and antero-posterior placement of the component was performed to optimize coverage. No medial tibial bone wear was encountered for patient's included in this study.

For both the kinematic and mechanically aligned knees, patella implantation was modelled as an onlay patella matching the resected surface at its posterior apex with an eight millimetres thickness patella button. The largest patella button that could fit on the resected surface without overhang was implanted and centred on the resected plane. The resection plane was drawn parallel to the patella tendon–quadriceps tendon attachment point axis and the femoral transepicondylar axis projected from the CT scan.

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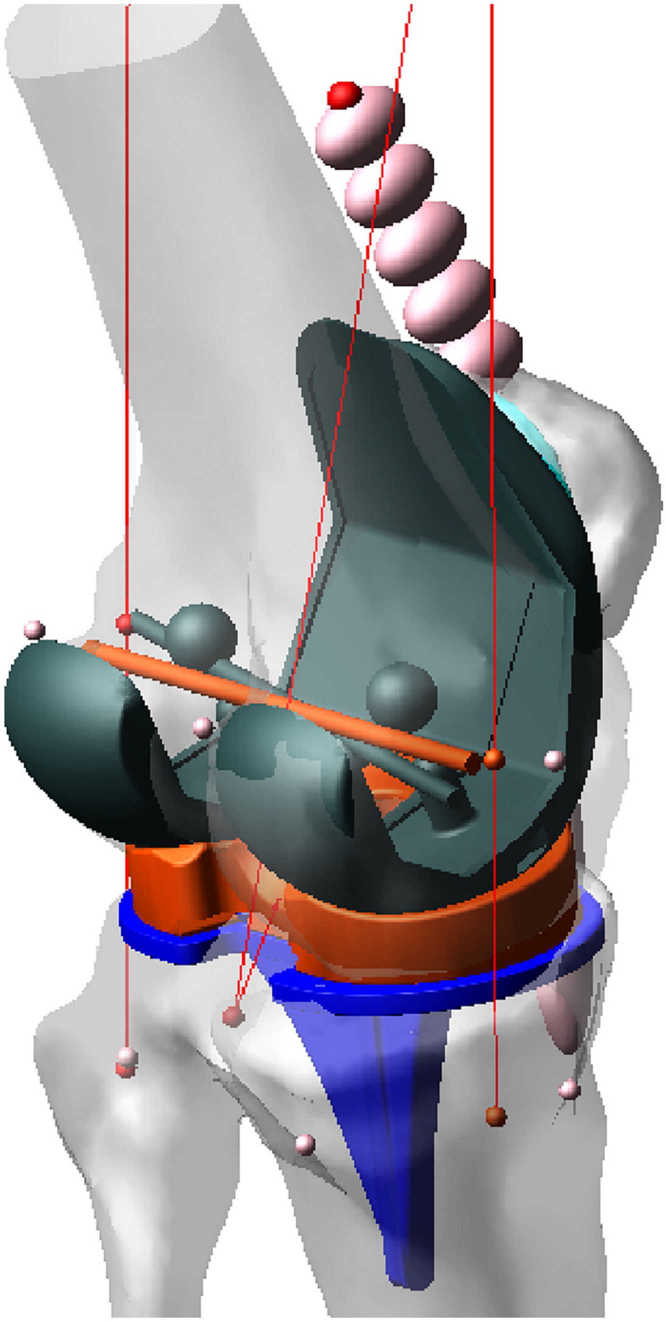

Figure 3. Simulation showing boundary conditions and ligaments present in the computational model (LCL, anterior MCL, posterior MCL, anterior PCL, posterior PCL). Ligaments were modelled as non-linear springs. Ligament forces were illustrated with the red lines. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

2.4. Data analysis

Kinematics was assessed using an implant to implant reference frame for both mechanical and kinematic alignment simulations and was based on the Grood–Suntay measurement system [37]. Reporting kinematics to bone based reference frames was trialled however the native mechanical axes results were dominated primarily by the static effects of component placement relative to the bone. Static placement of the implants in kinematic alignment was done independently.

Component placement for the kinematically aligned knees relative to the mechanical axes was then assessed. The simulated closed-kinetic-chain knee extension was performed and measurements of position were extracted. The medial and lateral flexion facet centre (FFC) condyles were identified as the point equidistant from the most distal and posterior planes of the implant, as

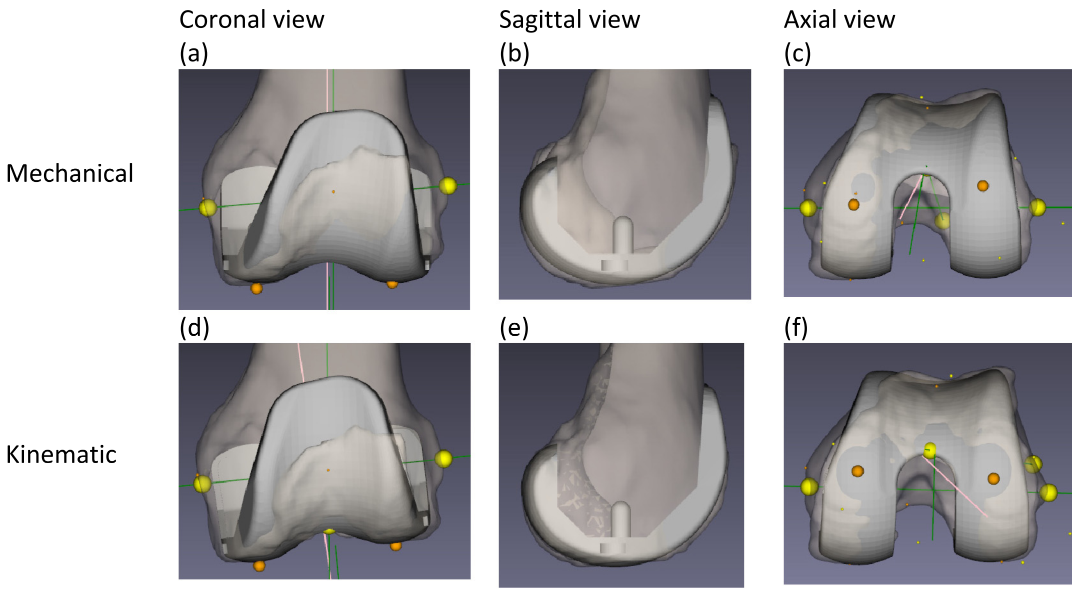

Figure 4. Mechanically aligned femoral component (a), (b) and (c); kinematically aligned femoral component (d), (e) and (f) in coronal, sagittal and axial views.

the multiradius implant design did not have a single flexion centre. These points were used as the reference points for measuring the movement of the femoral component relative to the tibia throughout the motion. The medial and lateral FFC measurements were taken from these reference points to the lowest dwell point on the tibial insert. Measurements were rescaled about the femoral AP measurement to account for implant size geometry.

Rollback was measured from the centre of these two FFC points to the tibial dwell point posterior translation of the transepicondylar line, hence is the average of the medial and lateral FFC translation measurements. The internal–external rotation measurement was the angle between the femoral and tibial components projected onto the tibial component plane. Patella lateral shift was defined as the translation from the centre of the patella button relative to the centre of the tibial insert, with positive in the lateral direction and negative in the medial direction. Patella external tilt was defined as external rotation of the patella relative to the transepicondylar axis of the femur projected onto the tibial plane.

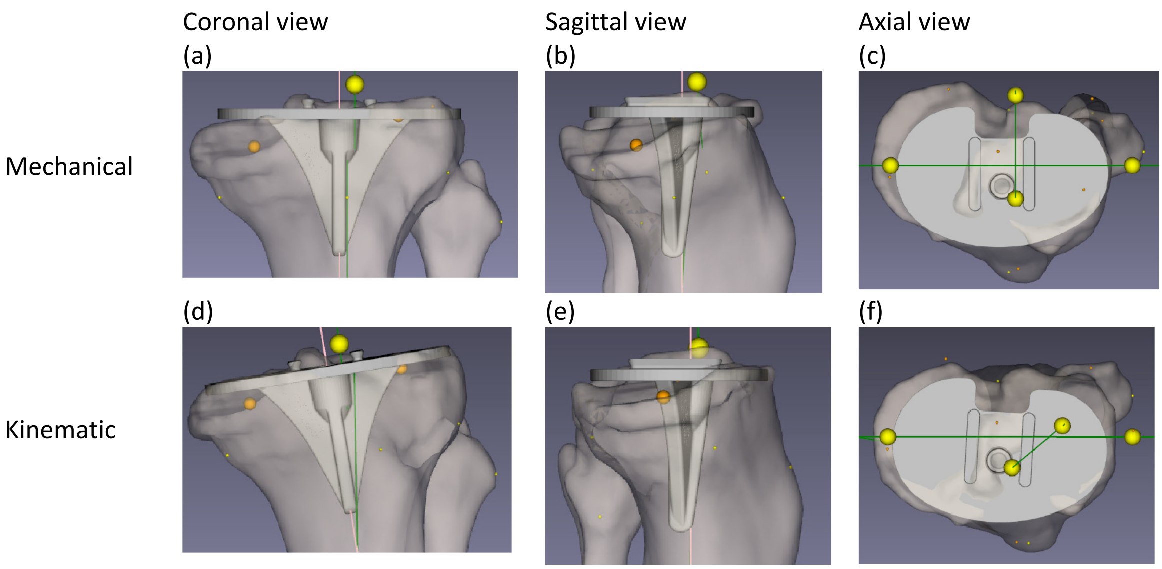

Figure 5. Mechanically aligned tibia component (a), (b) and (c); kinematically aligned tibia component (d), (e) and (f) in coronal, sagittal and axial views.

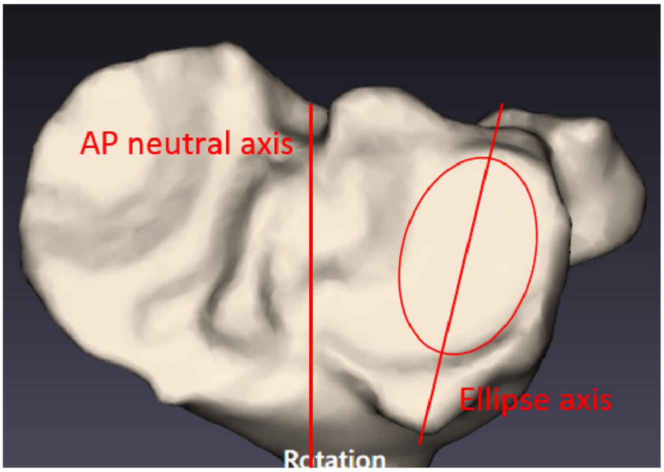

Figure 6. Ellipse used to define kinematic rotation angle and its angle relative to tibial AP rotational axis.

3. Results

3.1. Simulation component alignment for kinematically aligned knees

Native coronal alignment (hip–knee–ankle angle) for all knees as measured from CT scan had a mean of 3.1° ± 5.7° varus (range 8.7° valgus to 11.8° varus).

For mechanically aligned knees the femoral and tibial components were 0° to the mechanical axis. For kinematically aligned knees the mean tibial component coronal and axial alignment was 3.0° ± 2.4° (range −1.8° to 7.2°) varus to the mechanical axis and 7.2° ± 6.6° (range −9.4° to 15.4°) internal to tibial AP rotational axis respectively for kinematically aligned knees. Both component alignment parameter means were significantly different from mechanically aligned knees (0° varus and 0° rotation) ($p < 0.05$). Tibial slope in the kinematically aligned knees had a mean value of 4.6° ± 2.8° (range 0° to 11.2°). Kinematically aligned tibial slope mean was also statistically different to the mechanically aligned tibia slope (three degrees slope) ($p < 0.05$).

The mean femoral component coronal and axial alignment for kinematically aligned knees was 3.0° ± 2.3° (range −0.8° to 7.2°) valgus to the mechanical axis and 2.5° ± 1.6° (range −0.2° to 5.4°) internal to the surgical transepicondylar axis respectively. As with the tibial component placement, both component alignment parameter means were significantly different from mechanically aligned knees ($p < 0.05$). Femoral flexion in the kinematically aligned knees had a mean value of 2.4° ± 1.7° (range 0° to five degrees, as per

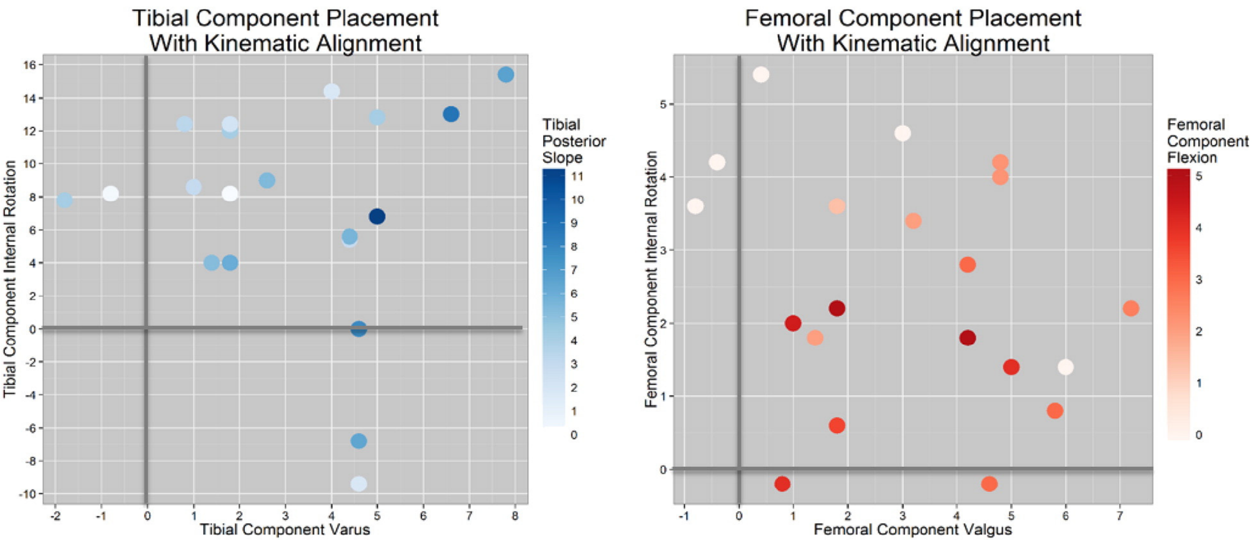

Figure 7. Coronal and axial component alignment for kinematically aligned knees. The cross section of the 0° horizontal and vertical axis represents mechanical alignment.

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the planning process). Femoral flexion mean in the mechanically aligned dataset had a mean value of 3.3 ± 1.7° (range 0° to five degrees) and was not statistically different to that of the kinematically aligned cases.

Figure 7 shows tibial and femoral component alignments for kinematically aligned knees. The horizontal and vertical lines represent 0° coronal and 0° axial alignment respectively. The cross section between the two lines is the mechanical alignment.

Figure 8 shows kinematic femoral and tibial component coronal alignment shaded by the native coronal alignment angle. The reference lines represent a three degree varus (blue), neutral (black) and three degree valgus (red) as the final coronal alignment. There was variation in the level of joint line obliquity with a given tibio-femoral coronal alignment. However, a trend between the native and kinematic tibio-femoral final alignment is present. Linear regression of final alignment as a function of native alignment yields an R^2 of 0.75.

3.2. Simulation tibio-femoral kinematics

Tibio-femoral kinematic results are shown in Figure 9. Statistically significant differences for paired t-tests at every time parameter were found ($p < 0.05$), with the exception of femoral AP translation from 30° of flexion and lower. The difference between the mean results for medial FFC AP translation for the kinematic and mechanical simulations starts at 0.4 ± 0.9 mm at five degrees flexion, increasing steadily to 1.7 ± 1.4 mm in deep flexion, kinematically aligned being anterior to mechanical. The lateral femoral FFC mean AP translation difference is 0.4 ± 0.6 mm at five degrees, increasing to 2.9 ± 1.9 mm in deep flexion with mechanically aligned anterior. The change in medial and lateral femoral FFC throughout flexion also implicitly describes the tibio-femoral internal–external rotation; kinematically aligned knees' lateral femoral FFC translates more posteriorly and medial femoral FFC translates more anteriorly than that of mechanically aligned knees', as flexion increases. Thus there is more external rotation of kinematically aligned knees. Also, there is relatively little difference in rollback behaviour, starting with no difference peaking at 0.8 ± 0.9 mm at 96°, kinematically aligned posterior to mechanical.

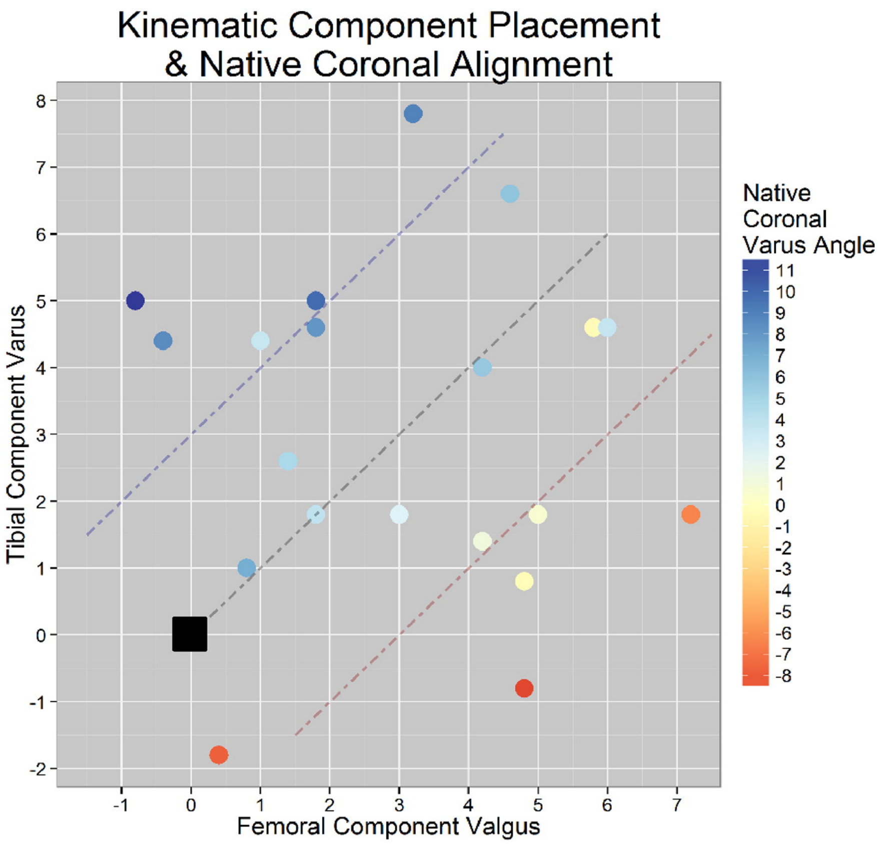

Figure 8. Kinematic alignment for femoral and tibial component valgus and varus angle shaded by the native coronal varus angle. The reference lines represent a three degree varus (blue), neutral (black) and three degree valgus (red) as the final coronal alignment. Mechanical alignment for femoral and tibial component is at zero (black square). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

3.3. Simulation patello-femoral kinematics

The patello-femoral kinematic results are shown in Figure 10. Patella lateral shift exhibited statistically different parameter values for measurements of flexion between 10 and 40° and at angles greater than 80°. Patella lateral shift for both alignment paradigms displayed a tendency towards medialising throughout the flexion cycle, though trend lines are different.

After starting in a common position, the kinematically aligned patellae tended towards shifting medially, peaking at 15° flexion where they were placed 1.8 ± 1.2 mm more medial. The kinematically aligned patellae then tracked without further medial lateral shift while the mechanically aligned continue to drift medially, finishing in deep flexion 2.2 ± 1.6 mm medial. Mean differences in patella lateral tilt under kinematic and mechanical alignments are significant up to 30° of flexion ($p = 0.05$). Kinematic alignment begins the simulation at 2.7° ± 2.1° more internal tilt relative to the mechanically aligned at five degrees flexion, with the kinematically aligned knees tilting internally by a mean 3.5° while the mechanically aligned knees are 0.8°. The means converge until about 60°, where they effectively show identical movement into five degrees external tilt at 140° flexion.

Intra-patient differences for patella tracking are high, however, with the difference in tilt for a given patient with either alignment approach ranging from six degrees more externally titled to 6.5° more internally tilted.

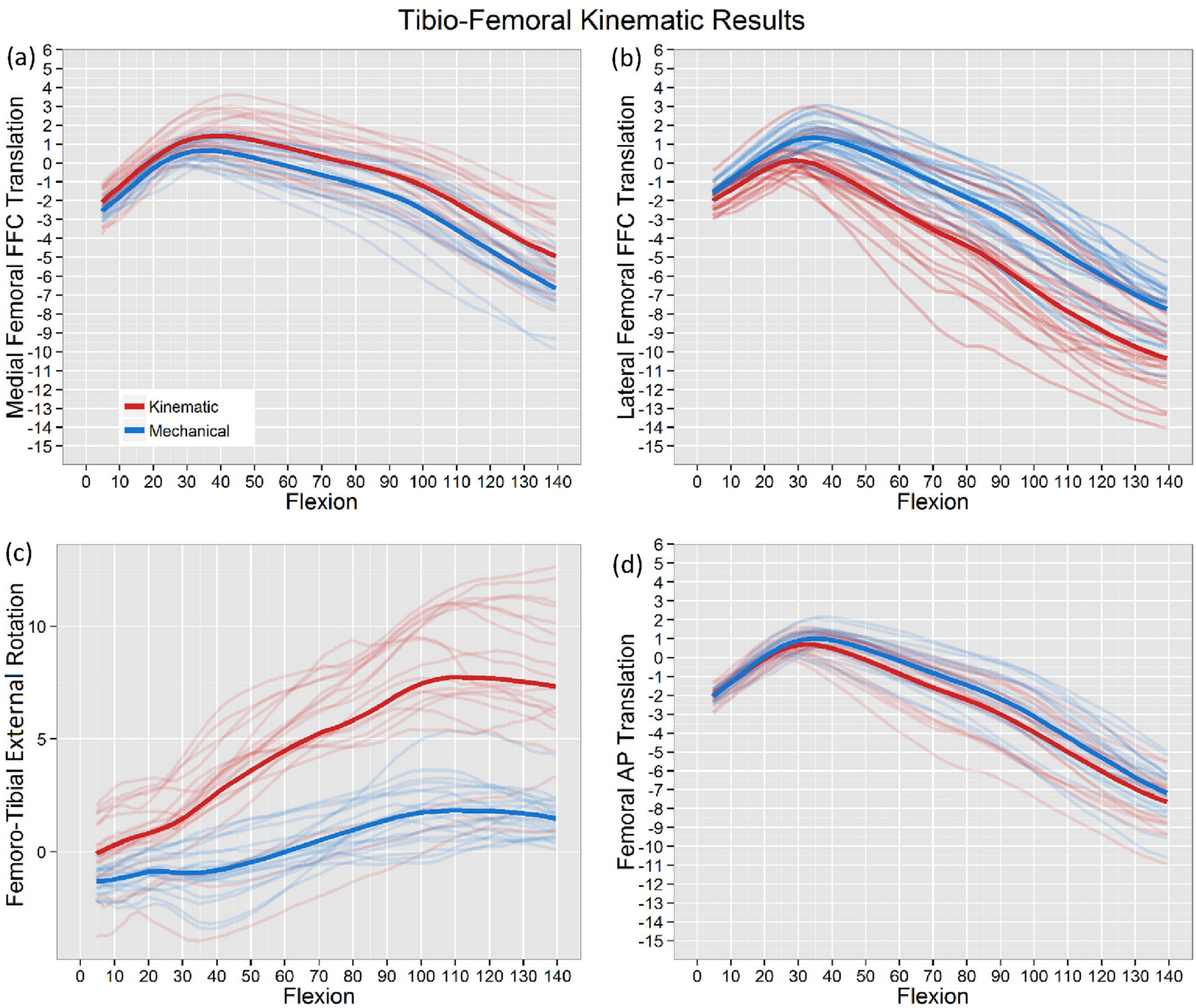

Figure 9. Tibio-femoral kinematic results for knee flexion. Red lines are kinematic alignment and blue lines are mechanical alignment. Solid lines are averages of each alignment. (a) and (b) medial and lateral flexion facet centre (FFC) antero-posterior drift from the lowest point of the tibial insert. Positive values indicate anterior translation. (c) Femoral–tibial internal external rotation. Positive values indicate external rotation. (d) Femoral AP translation relative to the lowest point of the tibia insert. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

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4. Discussion

Recent data has challenged the importance of traditional mechanical alignment philosophy [9]. Recently, Bellemans et al. [38] have introduced the concept of "constitutional varus", which hypothesizes that correction to a neutral mechanical alignment may not be "normal" for a significant proportion of the population. Their study showed that 32% of asymptomatic men and 17% of asymptomatic women possess a natural mechanical alignment of three degrees varus or more [38].

In conjunction with this principle, several surgeons have supported the restoration of kinematic, rather than mechanical, alignment in TKA [13,16,17] and Ji et al. [39] reported that native and 'healthy' joint line were one and the same for kinematically aligned knees. However, kinematically aligned knees show lack of consistency regarding patient outcomes, survivorship, and surgical technique [15–18,40,41]. Therefore, it remains unclear what are the optimal alignment targets for TKA despite of the emphasis on alignment philosophies for TKA.

Recently, Ishikawa et al. [27] used computational model to analyse the kinematics of kinematically aligned knees. Their study suggests that kinematically aligned knees produce near-normal knee kinematics. However, only a single model was used in the analysis and therefore the kinematics outcomes reported were limited.

In this study, pre-operative non-weight bearing CT scans of diseased joints in 20 patients were used to compare kinematic and mechanical alignments in a validated computational simulation. From patient CT scans, native coronal alignment was determined and kinematic and mechanical alignments were planned (Figure 8). Ishikawa et al. [27] used a clinically derived average kinematic alignment at three degrees tibial component varus and three degrees femoral component valgus which is equivalent to coordinates (3,3) on Figure 8. Our average alignment values were similar to reported alignment in clinical kinematic alignment studies [27]. However instead of using an average kinematic alignment, our study accounts for significant variation of patient pre-operative anatomy.

Results for kinematic alignment (Figure 8) showed that there was variation in the level of joint line obliquity with a set tibio-femoral coronal alignment. However, a trend between the native and kinematic tibio-femoral final alignment was observed. Any variation observed most likely occurred due to a condition of the pre-operative diseased joint and the wide range of adjustments necessary to attain kinematic alignment. When kinematically aligned, femoral components on average resulted in more valgus alignment to the mechanical axis and internally rotated to surgical transepicondylar axis whereas tibia component on average resulted in more varus alignment to the mechanical axis and internally rotated to the tibial AP rotational axis. This is consistent with other reports [16,18].

In regard to tibio-femoral kinematics, both kinematic and mechanical alignments resulted in a broad trend towards anterior translation of the femoral component up to 30° flexion, followed by posterior translation as flexion increases (Figure 9). The kinematically aligned knees experience external rotation of the femoral component on the tibial component during flexion, with the angle increasing steadily from 1.2° ± 1.5° at five degrees flexion to 5.9° ± 3.3° at 140° flexion. This internal rotation of the femur relative to the tibia as the knee reaches full extension is comparable to screw home mechanism observed in native knee motion [42]. This effect is less so for mechanically aligned knees.

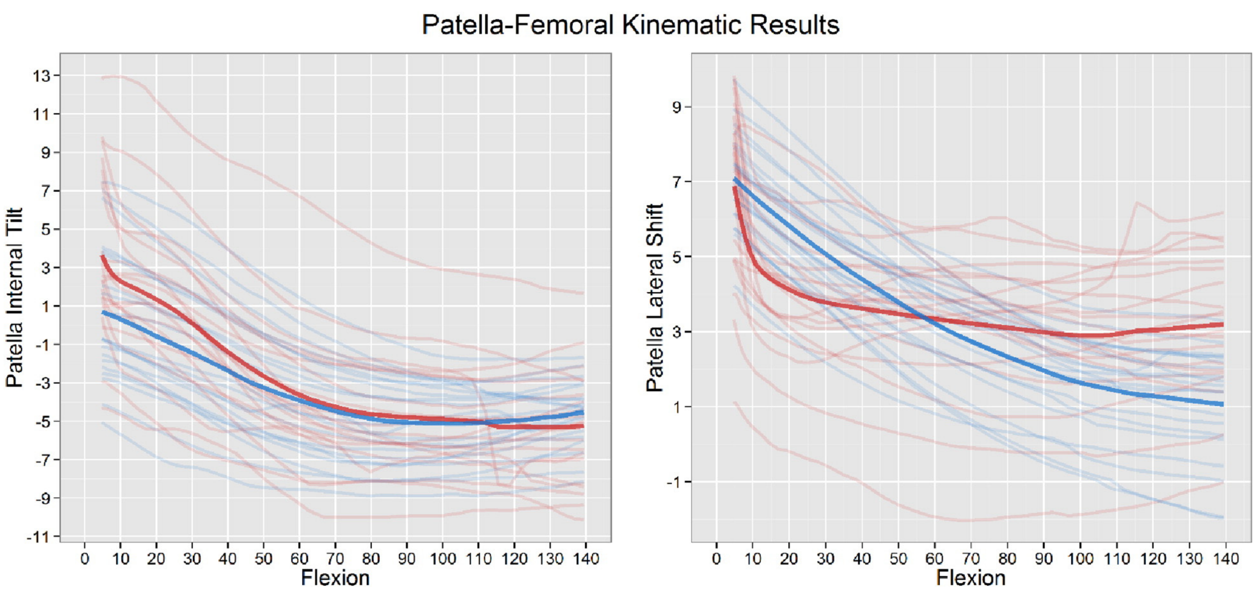

Figure 10. Patello-femoral kinematic results for patella external tilt (left) and patella lateral shift (right) for knee flexion. Red fine lines are kinematic alignment and blue fine lines are mechanical alignment. Solid lines are averages of each alignment. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

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In regard to patello-femoral kinematics, for both kinematic and mechanical alignments there were high intra-patient differences for patella tracking (Figure 10). However, the difference in tilt for a given patient with either mechanical or kinematic alignment ranged from six degrees more externally titled to 6.5° more internally tilted. There was less medial movement of the patella in deep flexion in kinematic aligned than mechanically aligned knees, though it arrived as its medial–lateral position earlier in the flexion. Differences in component alignment and potential impact on Q angle could explain some of the variation seen in patella-femoral kinematics for kinematically aligned knees.

Results for tibio-femoral kinematics for flexion and rotation as well as patello-femoral kinematics for tilt and shift were similar to that of previous computational biomechanical studies [19,24–27]. Variations existed primarily due to patient CT input, on which knee joint testing rig was simulated, e.g. Oxford Knee Rig or Kansas Knee Simulator, or if the implant was cruciate retaining (CR) or substituting (PS), or the alignment strategy simulated.

Our results for tibio-femoral kinematics for flexion and rotation using both mechanical and kinematic alignments closely match results reported by Ishikawa et al. [27]. All models exhibited anterior translation of the femoral component relative to the tibia during the early flexion phase and then posterior translation as flexion increased. The anterior translation from 0° to 30° of flexion was similar bilaterally in all models.

Patella lateral shift kinematics also replicated a similar pattern of mechanical alignment to that reported by Ishikawa et al. [27]. However, patella lateral shift kinematics for kinematic alignment as well as patellar external tilt for both alignments varied markedly between our results and those reported by Ishikawa et al. [27]. In the study reported by Ishikawa et al. [27], in the kinematic alignment models the patella tilted more externally relative to the femoral component at 0° and 30° and after 60° increased in all models. It was similar in our study until 60 to 70° and then tilting plateaued. Plateauing after 60° flexion was also reported by Kobayashi et al. [43] using healthy subjects in an in vivo study. Other explanations for this difference could be patient anatomy (one subject versus 20), model assumptions, patellar button size or design of the intercondylar notch and anterior patella groove of the femoral component.

There were several limitations in this study. The study involved 20 subjects only and this may be insufficient given how variable knee alignment is across the population. The implants used in this study were multi-radius femoral component with a single design fixed bearing cruciate retaining TKA. Therefore the results may not be applicable to other knee designs nor to mobile bearing or posterior stabilised knees. Also, the kinematics analysed were for closed-kinetic-chain knee extension and therefore functions such as walking or stair climbing may not be comparative.

The simulation model was subject to assumptions and variables common to many computational models: boundary conditions and muscle forces were assumed, only the lower extremity was modelled, there was limited soft tissue representation and cartilage was not accounted for. Such assumptions and variability are consistent with other computational modelling as well as in vitro modelling studies. However, computational modelling does offer the ability to simulate kinematics of different alignments on the same subject and thereby be potentially used as a predictive tool for pre-operative scenarios. Moreover, there are a number of studies that have shown that computational models could predict forces and kinematics that compared favourably to those found experimentally or in vivo fluoroscopy [19,24–26].

5. Conclusions

In conclusion, kinematic alignment had more variation than mechanical alignment for all tibio-femoral and patella-femoral kinematics. This was particularly true for tilt and shift of the patella-femoral joint for kinematically aligned knees. Kinematic alignment corrects long leg alignment to a patient specific alignment which depends on the preoperative state of the knee. Also, when kinematically aligned, femoral components on average resulted in more valgus alignment to the mechanical axis and internally rotated to surgical transepicondylar axis whereas tibia component on average resulted in more varus alignment to the mechanical axis and internally rotated to the tibial AP rotational axis. The use of computational models has the potential to predict which alignment, kinematic or mechanical, could improve knee function patient specifically.

Conflicts of interest

The authors declare that Willy Theodore, Joshua Twiggs, Elizabeth Kolos, Brad Miles, Justin Roe, Brett Fritsch, David Dickison, David Liu and Lucy Salmon were employed by or consultants to 360 Knee Systems.

Author contributions

WT, JT and BM conceived and designed the experiments; WT and JT performed the experiments; WT and JT analysed the data; BM, JR, BF, DD, DL, LS, and SH contributed materials/analysis tools; WT, JT, EK, BF and DL wrote the paper.

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Appendix A. Supplementary data

Supplementary data to this article can be found online at http://dx.doi.org/10.1016/j.knee.2017.04.002.

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