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Previous in vivo studies have observed that current designs of posterior stabilised (PS) total knee replacements (TKRs) may be ineffective in restoring normal kinematics in Late flexion. Computer-based models can prove a useful tool in improving PS knee replacement designs. This study investigates the accuracy of a two-dimensional (2D) sagittal plane model capable of predicting the functional sagittal plane kinematics of PS TKR implanted knees against direct in vivo measurement.

Implant constraints are often used as determinants of anterior–posterior tibio-femoral positioning. This allowed the use of a patello-femoral modelling approach to determine the effect of implant constraints. The model was executed using motion simulation software which uses the constraint force algorithm to achieve a solution. A group of 10 patients implanted with Scorpion PS implants were recruited and underwent fluoroscopic imaging of their knees. The fluoroscopic images were used to determine relative implant orientation using a three-dimensional reconstruction method. The determined relative tibiofemoral orientations were then input to the model. The model calculated the patella tendon angles (PTAs) which were then compared with those measured from the in vivo fluoroscopic images. There were no significant differences between the measured and calculated PTAs. The average root mean square error between measured and modelled ranged from 1.17\textdegree to 2.10\textdegree over the flexion range. A sagittal plane patello-femoral model could conceivably be used to predict the functional 2D kinematics of an implanted knee joint. This may prove particularly useful in optimising PS designs.

1. INTRODUCTION

Total knee replacements (TKR), whilst successful in providing pain relief, do not fully restore normal knee kinematics in many designs as shown by kinematic comparisons in the literature\cite{1,2,3}. This disparity is most apparent when patients attempt to perform activities that place large demands on the knee joint\cite{1,2,4}. It is thought that this is a result of the inability of TKRs to induce adequate femoral rollback with progressive knee flexion. Femoral rollback has the potential advantages of improved knee flexion, range of motion, and increased quadriceps’ moment arm.

Increasingly implant designs incorporate additional constraints to influence the kinematics of the replaced knee. The most frequently used additional constraint is that of the posterior stabilised total knee replacement designs. Posterior stabilised (PS) total knee replacement designs utilise a cam/post mechanism to generate femoral rollback in higher flexion. The concept of the cam/post mechanism was first introduced in the late 1970’s\cite{5} and is designed to replicate the function of the posterior cruciate ligament (PCL). Posterior stabilised TKR is intended to induce femoral rollback by mechanical interaction of the cam, incorporated within the femoral component, with the post extending from the tibial insert, guiding
the femur posteriorly along the tibia as knee flexion increases.

Though the effectiveness of the cam/post mechanism is frequently debated in the literature, clinical trials have not produced conclusive evidence supporting the excision or retention of the posterior cruciate ligament (6). Studies examining the kinematic performance of PS TKR designs(7–12) report observing femoral rollback. Fantozzi et al. (13) correlated significantly higher clinical and functional scores with a more posterior condylar position. Our previous studies (1,2) which used the patella tendon angle (PTA) to quantify kinematics, have observed that the PS knee may be ineffective in sufficiently restoring normal kinematics in higher flexion.

Design aspects of cam/post mechanisms such as point of engagement, increased posterior movement, or both, influence the resulting functional kinematics of a posterior stabilised total knee replacement. A number of methods have been used to assess kinematics post TKR, the most commonly used method being three-dimensional reconstruction of single plane fluoroscopic imaging(14–16). The ability to approximate the functional kinematics of a given PS TKR design would be useful in the process designing and redesigning of cam/post mechanisms. Computer based modelling methodologies can offer a suitable adjunct, providing a means of approximating the kinematics of PS designs without unnecessarily having to subject patients to inconvenience, tie up valuable imaging resources, and more importantly wait for designs to be implanted before testing them.

The cam/post mechanism, by design, is intended to influence anterior-posterior movement in the sagittal plane. The PTA reflects the effect the femoral anterior-posterior movement has on the patello-femoral interaction in the sagittal plane (Figure 1). PS TKR designs are more suited to modelling due to the relative anterior-posterior position of the femoral and tibial components being more predictable. This is based on the assumption that the cam/post mechanism engages resulting in a constrained relative anterior-posterior position of the femoral component to the tibial component. However, before cam/post engagement PS TKR knees, like cruciate retaining designs, are unconstrained or the constraints difficult to determine.

Models capable of approximating the functional kinematics of PS TKR implanted knees would aid in optimising future cam/post designs. Gill & O'Connor(17) proposed one such model which predicted the PTA for a relative anterior-posterior tibio-femoral relationship. However, little is known about how well their proposed model is able to approximate functional kinematics in vivo. This paper aims to investigate the accuracy of a sagittal plane modelling procedure (based on that by Gill & O'Connor(17)) developed to approximate the functional kinematics, based on the PTA, by comparing the model outcomes to those observed in vivo.
2. METHODS

The methodology is designed to directly compare the PTA calculated by the model to those measured from in vivo fluoroscopic images. As such the methodology used is best described as two separate parts. The first part describes the modelling approach and the second part the methodology of the validation of the model.

2.1 Part 1: The Modelling Approach

A two-dimensional modelling approach based on that proposed by Gill & O'Connor(17) was used. The aim of the model was to approximate the functional kinematics of a posterior stabilised knee replacement. As a measure of functional kinematics the PTA was used. The PTA is the angle subtended between the patella tendon and the long axis of the tibia. As described in the introduction, the PTA is directly influenced by the position of the femur on the tibia, as a component of the extensor mechanism (Figure 1). Specifically, the model aimed to determine the relationship of the PTA with the knee flexion angle (KFA) resulting from a given tibio-femoral position.

The driving input to the model was the relative position of the femoral component to that of the tibial component of the implant. Taking the tibio-femoral position (as would be determined by cam/post interaction in PS TKR designs) as the driving input, the resulting PTA was calculated using a two-dimensional patello-femoral model based on that developed by Gill & O'Connor(17). In addition, a number of anatomical parameters are required.

When calculating the PTA, it is important to recognise that anatomical differences have an influence on the PTA/KFA relationship. To accommodate this, the model was patient specific; anatomical parameters for each patient were determined individually and supplied as additional inputs to the model.

2.1.1 Input Parameters

In addition to the tibio-femoral position a number of parameters describing anatomical constraints were required in the modelling procedure. These parameters are illustrated in Figure 2 and defined below:

- **Position of Tibial Component.** The proximal anterior point of the tibial component was taken as the origin of the global reference axis relative to which the other points in the model were determined.
• **Position of Tibial Tubercle (T).** The tibial tubercle is defined as the point where the patella tendon attaches to the tibia (*Figure 2*). The position was recorded relative to the tibial component.

• **Patella Tendon Length (PTL).** The patella tendon length was calculated as the distance between the tibial tubercle (T) and the distal pole of the patella (P_d), where the patella tendon attaches to the patella (*Figure 2*). The patella tendon attachment on the patella covers a small region of the patella; the centre of this region was taken as the attachment point for the purpose of this model.

• **Patella Length (PL).** The patella length is the distance between the distal (P_d) and proximal (P_p) poles of the patella (*Figure 2*).

• **Patella Thickness (P_th).** The patella thickness was defined as follows: a line was constructed between the P_d and P_p and the perpendicular distance between this and the closest point on the femoral component was taken as the patella thickness.

• **Quadriceps Force (QF).** As an equilibrium type modelling approach is used an input quadriceps force was required. The force applied as the quadriceps force in the model was based on the approximately linear quadriceps force/KFA relationship measured by Bowne *et al.* (18). The quadriceps force value was calculated as a linear interpolation of the line between 200N at 0° KFA and 900N at 90°.

**2.1.2 Component Geometry**

The geometry of the components was defined using the drawing exchange format (DXF) which could easily be edited in, and exported from, most CAD packages. The majority of implants are now designed in three-dimensions (3D). The model was two-dimensional (2D) which required the 3D CAD information to be converted to useable 2D information. The sagittal outline of the implant was used in the model for both the tibial and femoral components. For the femoral component the sagittal outline used was that of the contact surfaces, that is the average outline of the condyles and the trochlea groove.

**2.1.3 Calculation of the Patella Tendon Angle**

The driving input of the patello-femoral model was the femoral position relative to the tibia. The tibia was considered to be fixed. The femoral position was defined by the position of the origin of the femoral coordinate system given as an x and a y coordinate and a rotation angle µ. The x coordinate defined the anterior-posterior position, the y coordinate the distal-proximal position, and µ described the rotation of the component.
The model of the patello-femoral joint used was based on that proposed by Gill & O’Connor(17). A number of assumptions were made:

- The difference in thickness between the trochlea groove and the patella facet contact surfaces on the femoral condyles of the implant modelled in this paper was assumed negligible. This allowed the patella to be represented as being rectangular in shape as proposed by Yamaguchi with a single contact surface, which could interact with either the trochlea or femoral condyles.

- The patella tendon was considered to be inextensible, and the patella tendon was pin jointed to the patella distal pole and the tibial tubercle.

- The force acting through the quadriceps tendon was parallel to the femoral axis until quadriceps tendon wrap took place.

- Both the patella and quadriceps tendons were in tension at all times during the step up and lunge exercises.

- The patella tracks along the given geometry as intended by the TKR design.

- Gill & O’Connor modelled the contact as being frictionless. In this model friction was accounted for; the values for a cobalt-chrome/polyethylene bearing couple were used. The friction coefficient used to describe the contact between the patella and femoral component was 0.2.

Using the above constraints enabled the movement of the patella relative to the femur to be determined. The correct position of the patella corresponding to a particular tibio-femoral relationship was found by assuming that the forces through the patella were in equilibrium at any given moment. The point of equilibrium was calculated using motion simulation software (Working Model, version 7.0.0.0, MSC. Software Corp, Redwood City, CA, USA) which uses a constraint force algorithm(19) to achieve a solution. The output PTA value was the value of the PTA when the model was in equilibrium.

At higher degrees of flexion, quadriceps tendon wrap needed to be accounted for. Quadriceps tendon wrap occurs when the quadriceps tendon comes into contact with the trochlea causing it to wrap around the femur in high knee flexion angles. Wrapping results in the quadriceps tendon no longer being parallel to the femoral axis. The model accounted for this by keeping the orientation of the quadriceps tension unchanged beyond the point at which the quadriceps tendon wrap occurred. Quadriceps tendon wrap was taken to occur at 87.5° of KFA, as reported by Gill & O’Connor(17).
2.2 Part 2: Model-in vivo comparison

To determine whether the patello-femoral model accurately approximated the PTA resulting from the relative tibio-femoral position throughout the range of flexion, the tibio-femoral positions obtained from *in vivo* fluoroscopic data were input into the model, and the model calculated PTA values were compared to the corresponding PTA values measured directly from the same *in vivo* fluoroscopic data using a previously developed methodology (2,20).

The input parameters (described in 2.1.1) for each individual patient were obtained from *in vivo* recorded fluoroscopic data (2). To measure the parameters from the *in vivo* data, the fluoroscopic images were corrected for distortion (20–22), and a user interface developed in Matlab (version 7, The Maths Works Inc, Natick, Massachusetts) then allowed the anatomical landmarks to be interactively templated for each image from any given sequence of recorded data. For each image, values for the parameter set were calculated and an average of these values over the entire image sequence was used in the model. The parameters templated are illustrated in Figure 2 and defined below:

A group of ten patients (6 females, 4 males) who had received a Scorpio PS (Stryker, Newbury, UK) knee arthroplasty at least twelve months prior to assessment were recruited. The patella was resurfaced in every case, the average age of the PS group was 70.3 years (range: 64 – 78 years). All patients had either excellent or good scores according to the American Knee Society Scoring system [average objective score: 90 (range: 86 – 100), average functional score: 95 (range: 90 – 100)]. The patients were asked to perform a step-up exercise and a deep knee bend whilst undergoing fluoroscopic imaging. Kinematic data were obtained using a previously developed standard fluoroscopic technique (1). Images were sampled at 25 frames per second, ensuring that the knee remained in the fluoroscopy field throughout the exercise. For each recording, the plane of the fluoroscopic image was aligned parallel to the sagittal plane of the knee.

For each patient, the individual anatomical parameters were determined as described above. The two-dimensional geometrical descriptions were created from the appropriate three dimensional CAD description provided by the manufacturer (Stryker, Newbury, UK). The input (tibio-femoral position) for each KFA was determined using a previously developed 2D/3D reconstruction method (2,20). A projection matrix, unique for each patient session, was calculated mapping the 3D coordinates of the measurement volume to the 2D coordinates of the distortion-corrected image. Using geometric models (CAD models) of the implant components, placed in the virtual measurement volume, it was then possible to determine the movement of the femoral component relative to the tibial component throughout the image sequence and thus throughout the knee flexion range. The patello-femoral model was then run using the patient specific parameters and
the corresponding measured tibio-femoral position determined *in vivo* as inputs, and the calculated
PTA/KFA relationship obtained as an output (in this paper these output results are referred to as the "*model calculated PTA*").

The model calculated PTA/KFA relationships were then compared to those actually measured from the
fluoroscopic data using a previously developed methodology(2,20); this was termed the "*measured PTA*".
For the whole data set, the average values of calculated PTA were compared to the average values of the
measured PTA at every 10° knee flexion interval using Student’s t-test (analysis performed using Matlab,
Version 7.0.0 (R14), The Maths Works Inc., Natick, Massachusetts). Error values were obtained by
subtracting the model calculated PTA values from the measured PTA values for each individual data set.
Variation between the model calculated PTA data and the measured PTA data was further explored using a
root mean square error (RMSE) analysis. The correlation between model calculated PTA data and
measured PTA data was explored using Spearman’s rank correlation coefficient. The agreement between
the model’s results and the measured values were assessed using a Bland-Altman plot(23).

The sensitivity of the model to variation in the input parameters was also investigated. The reliability of the
templating method was determined by asking an experienced surgeon to template the parameters for a
single patient ten consecutive times, with each occasion separated by at least an hour. The standard
deviation (SD) of each parameter measurement was determined. The model’s sensitivity to the measured
variations was investigated by considering variations to each input parameter of ±1.96SD (i.e. the 95%
confidence intervals) and running the model. The effect of the parameter variation was quantified by
subtracting the PTA values calculated using the original parameters from the PTA values calculated using
each parameter adjusted by ±1.96SD. These values were calculated over the knee flexion range and then
averaged to give a mean error.
3. RESULTS

The mean measured PTA ranged from 11° in extension to -1° at 100° of knee flexion; the model calculated PTA was similar to the measured PTA ranging from 10° in extension to 0° at 100° of knee flexion (Figure 3). A relatively large variation was present for both the measured and model calculated data sets. This was a result of the large anatomical differences between individual patients. The results of the Student’s t-test showed that there were no significant differences between the measured and model calculated PTAs at any point in the range (Table 1). The error values at 10° intervals of KFA ranged between 1.09° and -1.23° and are represented in the form of a box and whisker plot in Figure 4. The average root mean square error (RMSE) in PTA ranged between 0.59° and 1.90° over the flexion range. Table 1 gives the average RMSE and its standard deviation as well as the significance (p-value) at 10° intervals.

A simple plot of the model calculated PTA values plotted against the measured PTA values (Figure 5) shows that the model was able to approximate PTA reliably. The Spearman’s rank correlation coefficient between the model calculated PTA and measured PTA values ranged from 0.72 to 0.98 and the correlations were highly significant with p values less than 0.05 for all the patients (Table 2). The Bland-Altman plot (Figure 6) showing the relationship between the difference and the mean showed that all the differences were within 1.96 standard deviations (SD) and that there was no bias.

The standard deviation of the parameters templated ranged from 0.87 mm to 1.74 mm (Table 3). The effect of the parameter variation ranged from -0.13° to -3.43° average error in PTA values (Table 3). The largest average errors were seen for the patella thickness parameter with values of -3.43° and 2.75° average error in PTA at -1.96SD and +1.96SD respectively. The effect of of the parameter variation for the remaining templated original parameters are also shown in Table 3.
DISCUSSION

A patello-femoral model was developed to calculate the sagittal plane functional kinematics of constrained total knee implants. The model was validated by comparing the PTA measured directly from in vivo patient data to PTAs calculated by the model, using actual in vivo anterior-posterior femoral positions as inputs.

The average PTA/KFA relationship calculated by the model corresponds to that measured directly from the in vivo data (Figure 3). Both the measured and calculated PTA values have a large standard deviation. This can be attributed to the anatomical differences between individual patients. A more detailed look at the error between model calculated and measured data for each individual data set was analysed using the differences in PTA between the two. The model calculated PTA for a particular KFA valued was subtracted from the measured PTA value at the same KFA to determine the difference. The average errors calculated (Figure 4) varied between 1.09° and −1.23°, with a standard deviation of 2.00°. The errors showed no noticeable trend, such as an increase or decrease of error with flexion.

In investigating whether the model accurately approximates the PTA/KFA relationship of the knees, it is also possible to look at the agreement between the two sets of data. Essentially the two data sets can be seen as two methods of measuring the PTA, as opposed to assuming the measured PTA to be the gold standard and comparing the model calculated PTA to it. If the model calculated and measured values agreed perfectly, a plot of all the modelled values against the measured values would give a series of points lying on the line \( y = x \). In the case of this experiment the points do not all lie on the line \( y = x \), but are closely clustered about the line (Figure 5). The distance of the points from the line indicates the difference between the values determined by the different methods. The clustering of the points about \( y = x \) in Figure 5 indicates good agreement between the two methods. A more informative manner of assessing the agreement of the two data sets is to plot the difference between the two methods against their mean (Figure 6). Assuming the differences are normally distributed, the plot shows that 95% of the model calculated PTAs are within ±3.5°. The differences are likely to be normally distributed as the variations between subjects are removed by measuring the difference against the mean, indicating that there is a relationship between the two methods (Figure 5). The plot shows that the agreement of the two methods is acceptable with 95% of values within 3.5° of each other. Additionally, there was also no clear bias towards either method.

The patient specific nature of the model meant that the outcome of the model closely represents the kinematic profile seen in a given patient. A patient specific model allows the approximation of the effect design alterations to a PS TKR would have on a particular patient. In particular it allows the model to take into account implant sizes. In particular it allows the model to take into account implant sizes. The determination of the patient specific parameters does bring with it certain limitations. The parameters
templated from radiographic images introduced a certain amount of error. These errors were investigated in this study, as were the possible effects of such errors on the model’s output. The greatest error was introduced by the variations in the patella thickness (Table 3). The remaining parameters influenced the calculated PTA markedly less than the patella thickness did. The influence of the patella tendon length on the calculated PTA in particular was significantly less, meaning that modelling it as inextensible was a reasonable assumption.

The main limitation of the model was its two-dimensional nature. The movement of the patella in the knee is three-dimensional, moving mediolaterally as well as in the sagittal plane. Due to the two-dimensional nature of the model the effects of certain parameters that may affect the PTA are not taken into account. The most prominent one being the relative internal and external rotation of the femoral component relative to the tibia. In addition to this, the manner in which the patello-femoral contact was modelled differed from the real complex interaction of the condylar facets, trochlea groove, patella facets and patella prominence. The representation of the patella as a rectangle, further simplifying the model used by Gill et al.(17), was considered sufficient as it would be difficult to determine the patient specific differences between the different contact surfaces, particularly in cases where the patella is not resurfaced. Additionally the method used to account for wrapping of the quadriceps tendon assumes that the trochlear groove has a constant radius in the sagittal plane introducing potential for error. The two-dimensional nature of the model also meant that relative tibio-femoral rotations are not taken into account. Although the effect of the rotations on the PTA would be seen in the output, it would not be possible to distinguish between the influence of the rotation or genuine anterior-posterior movement. Although rotation and AP movement cannot be distinguished separately, AP movement does account for the greater part of the tibio-femoral joint's influence on the PTA. This, and the fact that the majority of the knee joint’s movement occurs in the sagittal plane, meant the model was effective in approximating the influence of the tibio-femoral trends on the PTA/KFA relationship.

The PTA has been used in the past by several authors(24–29) when studying knee kinematics of TKRs, unicompartmental replacements, patello-femoral replacements, as well as asymptomatic knees. It is directly influenced by both the tibio-femoral joint and the patello-femoral joint (Figure 1), as the orientation of it affects both the direction and magnitude of the forces in the extensor mechanism. The aim of the posterior stabilised knee is to restore the relative tibio-femoral position to normal. However, this may be achieved without the extensor mechanism performing as it would normally. The force direction, magnitude and patella tendon moment arm may be altered compared to normal knees(2). From this observation it could be concluded that the PS knee fails to produce the optimal relative tibio-femoral position. A model able to calculate the influence of the tibio-femoral component relationship on the gross kinematics of the knee would prove helpful in investigating the effectiveness of particular design changes. In posterior stabilising
TKR implants the anterior-posterior position of the femur on the tibia is constrained by the cam/post mechanism once it has engaged. This allows for the theoretical tibio-femoral position to be determined for knee flexion angles at which cam and post are engaged giving an input to the proposed model. Taking this a step further the model can be used to optimise the shape of either or both the cam and post so as to achieve an optimal patella tendon angle and thus optimal sagittal plane kinematics. The use of the described model to improve cam/post design is best illustrated using a brief example. If the PTA of a particular design in at 90° of knee flexion were found to be X° higher than that observed in the natural knee (assuming natural knee kinematics are desired) it would be reasonable to assume that placing the post Y mm more posteriorly relative to the tibia would force the femoral component posteriorly lowering the PTA. However, the question remains by how much would the post position Y need to be adjusted to reduce the PTA by X°. The described model allows for the effect of relative tibiofemoral change on PTA to be approximated which enables the optimization of the cam/post mechanism.

In conclusion the two-dimensional model in this study was shown to accurately approximate the patient specific PTA/KFA relationship resulting from a relative tibio-femoral position. This, combined with in vivo measured data set of anatomic parameters and tibio-femoral motion, could provide a useful tool in future analyses of posterior stabilised knee designs, in particular studies focussing on design optimisation.


### Tables

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<th>KFA (degrees)</th>
<th>0</th>
<th>10</th>
<th>20</th>
<th>30</th>
<th>40</th>
<th>50</th>
<th>60</th>
<th>70</th>
<th>80</th>
<th>90</th>
<th>100</th>
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<tbody>
<tr>
<td>Average RMSE</td>
<td>1.51</td>
<td>1.23</td>
<td>1.57</td>
<td>1.16</td>
<td>1.37</td>
<td>1.89</td>
<td>1.9</td>
<td>1.59</td>
<td>0.96</td>
<td>1.46</td>
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<td>1.1</td>
<td>1.05</td>
<td>1.02</td>
<td>1.42</td>
<td>1.32</td>
<td>1.91</td>
<td>0.99</td>
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<td>p-value</td>
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<td>0.52</td>
<td>0.27</td>
<td>0.42</td>
<td>0.59</td>
<td>0.37</td>
<td>0.5</td>
<td>0.39</td>
<td>0.39</td>
<td>0.91</td>
<td>0.10</td>
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**Table 1:** Table showing RMSE, RMSE STD and significance at 10° intervals.

<table>
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<tr>
<th>Patient</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
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<th>7</th>
<th>8</th>
<th>9</th>
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<tr>
<td>Spearman's Corr Coef</td>
<td>0.89</td>
<td>0.9</td>
<td>0.65</td>
<td>0.90</td>
<td>0.74</td>
<td>0.98</td>
<td>0.90</td>
<td>0.72</td>
<td>0.90</td>
<td>0.93</td>
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<tr>
<td>p-value</td>
<td>&lt;0.05</td>
<td>&lt;0.05</td>
<td>0.05</td>
<td>&lt;0.05</td>
<td>&lt;0.05</td>
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<td>&lt;0.05</td>
<td>0.04</td>
<td>0.01</td>
<td>0.02</td>
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</table>

**Table 2:** Table showing Spearman’s correlation coefficient and significance of the correlation between measured and calculated data at 10° intervals for the individual patients.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>PL</th>
<th>Pth</th>
<th>PTL</th>
<th>Tx</th>
<th>Ty</th>
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<tr>
<td>SD of Templated Values $(mm)$</td>
<td>1.42</td>
<td>1.65</td>
<td>1.74</td>
<td>0.87</td>
<td>1.61</td>
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<tr>
<td>Average Error + 1.96SD $(degrees)$</td>
<td>0.63</td>
<td>2.75</td>
<td>-0.71</td>
<td>1.59</td>
<td>0.20</td>
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<tr>
<td>SD Error + 1.96SD $(degrees)$</td>
<td>1.43</td>
<td>1.49</td>
<td>1.14</td>
<td>0.98</td>
<td>1.31</td>
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<tr>
<td>Average Error - 1.96SD $(degrees)$</td>
<td>-0.74</td>
<td>-3.43</td>
<td>0.50</td>
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<td>0.82</td>
<td>0.77</td>
<td>0.83</td>
<td>0.82</td>
<td>0.87</td>
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**Table 3:** Table showing the standard deviation of the templated original parameters (SD of Templated Values). The average error in PTA (Average Error + 1.96SD & Average Error - 1.96SD), calculated by subtracting the PTA values of the model run with the original parameters from the values from the model run with the original parameters adjusted by ±1.96SD. The standard deviation of error (SD Error + 1.96SD & SD Error - 1.96SD) is also given for every parameter.
**Figures**

**Figure 1**: Illustration of the resulting change in PTA with a shift in anterior posterior position between axes $O_1$ and $O_2$ relative to the tibial axis $O_t$.

**Figure 2**: Illustration showing the anatomical parameters measured from fluoroscopic images - Patella Tendon Angle (PTA), Knee Flexion Angle (KFA), Tibial Tubercle position ($T$), Distal Pole Patella ($P_d$), Proximal Pole Patella ($P_p$), Patella Tendon Length ($PTL$), Patella Length ($PL$), Patella Thickness ($P_{th}$).

**Figure 3**: Graph showing the average PTA plotted against the KFA for both measured and model calculated data sets for all ten patients.

**Figure 4**: The errors between the model calculated PTA and measured PTA values, plotted at every 10° of KFA, over the range 0° to 100° KFA.

**Figure 5**: Graph showing the model calculated PTA plotted against the measured PTA on the same plot as the line $y = x$. $R^2 = 0.84$

**Figure 6**: Bland-Altman plot showing the difference in PTA (measured PTA - model calculated PTA) plotted against their mean.