



# THE INFLUENCE OF CONTEMPORARY KNEE DESIGN ON HIGH FLEXION IV: *A KINEMATIC COMPARISON WITH THE HEALTHY INTACT KNEE*

Orthopaedic Research Laboratories  
Cleveland, Ohio

Edward A. Morra, M.S.M.E.  
A. Seth Greenwald, D.Phil.(Oxon)

## INTRODUCTION

Although Total Knee Arthroplasty (TKA) surgery enjoys 90% of outcomes with good to excellent results, some patients are uncomfortable adjusting their gait to accommodate the new articulations inherent in many contemporary implant designs. Paradoxical motions, inclusive of anterior sliding and lateral pivot of the femur relative to the tibia are examples of aberrant TKA kinematics that are opposite of those observed in healthy intact knees.

A computational kinematic simulator is employed in this study to quantify the motion of six posterior stabilized TKA designs during high flexion activity, allowing comparison to the motion of healthy intact knees. The VEGA (Aesculap), Vanguard PS (Biomet), Apex PS (OMNI life science, Inc.), Journey II (Smith & Nephew), Legacy LPS-Flex Fixed and Persona PS (both from Zimmer) were evaluated. All six designs are fixed plateau and currently available for clinical use in the United States.

## COMPUTATIONAL KINEMATICS

KneeSIM, a dynamic, validated musculoskeletal modeling system was utilized in this study. It provides a musculoskeletal modeling environment of the left leg of a nominal sized patient in which activities such as walking gait, lunge, stair ascent and descent and deep knee bend may be simulated. Activities are propelled by muscle forces and constrained by soft tissues.

Solid models of TKA component geometries are arranged in the joint space to reflect a successful virtual surgery (Figure 1). A specified activity is simulated and animations and plots of component and soft tissue positions, forces and moments are generated.

Factors influencing kinematic function and stability of the knee joint, including surgical technique, component placement, design, and soft tissue competency may be varied within the KneeSIM modeling environment. Patient anthropometrics may also be varied.

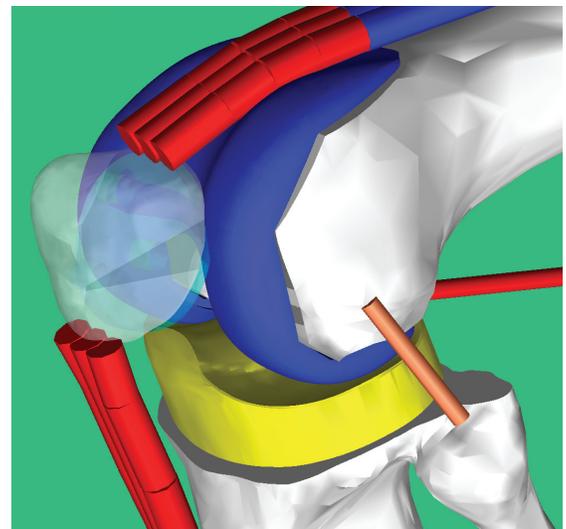


Figure 1: KneeSIM, a dynamic, validated musculoskeletal modeling system.

## CLINICAL EVIDENCE

The results from several video fluoroscopy studies, inclusive of the Duracon<sup>1</sup> and Journey BCS<sup>12</sup> knees have demonstrated that in vivo component motions for a given activity are very similar to those results predicted by the KneeSIM computational model. Patient specific KneeSIM models<sup>8</sup> have demonstrated the importance of bony impingement and its effect on patient flexion, and that the model closely predicts clinical weight bearing range of motion. In this study, impingement of the posterior femoral bone cut surface (Figure 2a) with the tibial insert (Figure 2b) defines the maximum flexion angle. The table below compares the maximum flexion angle predicted in previous KneeSIM models with values reported from clinical studies.

TKA DESIGN NAME	POSTERIOR STABILIZED?	MAXIMUM FLEXION ANGLE (degrees)	
		MODEL PREDICTION	CLINICAL AVERAGE
Triathlon	No	104	108 <sup>1</sup>
MRK	No	104	105 <sup>7</sup>
Duracon	No	105	105 <sup>11</sup>
Vanguard	Yes	117	111 <sup>6</sup>
Journey BCS	Yes	139	118 <sup>5</sup>
Legacy LPS-Flex Fixed	Yes	144	135 <sup>4</sup>

Model predictions for non-posterior stabilized designs are very similar to clinical results. The maximum clinical flexion angles reported for the posterior stabilized designs are less than model predictions, possibly due to limited pre-operative range of motion.

## STUDY METHODS

Three-dimensional solid models of the femoral, patellar and tibial insert components were “implanted” in the KneeSIM joint space per each manufacturer’s unique surgical procedure. Both cruciate ligaments were virtually resected in all six cases studied. Weight bearing high flexion activity was simulated, and femoral component motion was quantified as a function of knee flexion angle.

To aid in comparison with published weight bearing, healthy intact knee motion data<sup>3</sup>, a common marker describing femoral motion was required. In a manner similar to the clinical study, unique flexion facet centers (FFC) markers were determined for each femoral component using computer aided design tools (Figure 3). A sagittal plane was cut through each femoral condyle and a circle approximating the posterior condyle articulating surface was created. The FFC is depicted as a sphere at the middle of the circle, acting as a center of rotation through most of the flexion arc of motion. Medial and lateral flexion facet centers were joined to create a “barbell” structure, which was rigidly affixed to the femoral component to better visualize its motion.



Figure 2a: Posterior femoral bone cut surface.



Figure 2b: Maximum flexion defined by bony impingement.

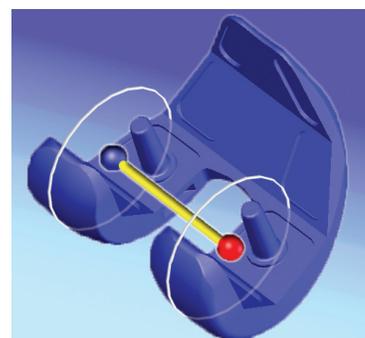


Figure 3: Determining flexion facet centers.

## RESULTS

The resulting animations and plots characterize motion of the femoral component relative to the tibial insert in comparison to that of the healthy intact knee. Each design flexes until the posterior femoral bone cut surface impinges against the tibial insert, then returns to full extension. Figures 4a, 4b, 4c and Figures 5a, 5b, 5c represent the moment when maximum flexion occurred for each design. The plot on the left illustrates anterior (positive values) and posterior (negative values) translation of the flexion facet centers as a function

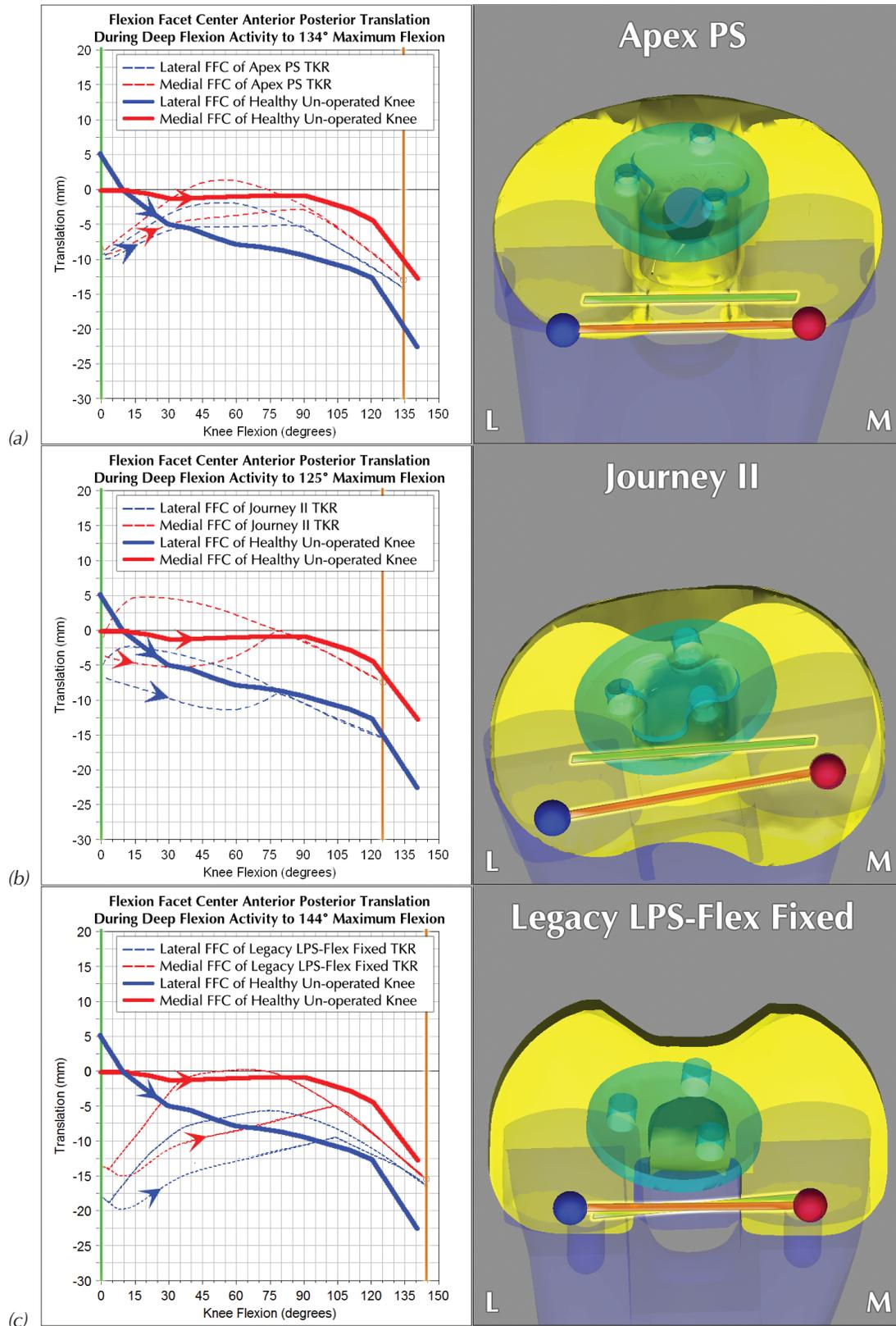


Figure 4

of knee flexion angle, with zero representing the midline of the tibial insert. The image on the right depicts component orientation at maximum flexion appreciated from a superior view. The blue sphere represents the location of the lateral FFC, and the red sphere the location of the medial FFC. Initial location of the FFC barbell at zero degrees of knee flexion is marked as a green bar, the location of the FFC barbell at maximum flexion is marked as an orange bar. These reference points contribute to understanding the relative motion of the femoral component. Designs are presented in alphabetical order.

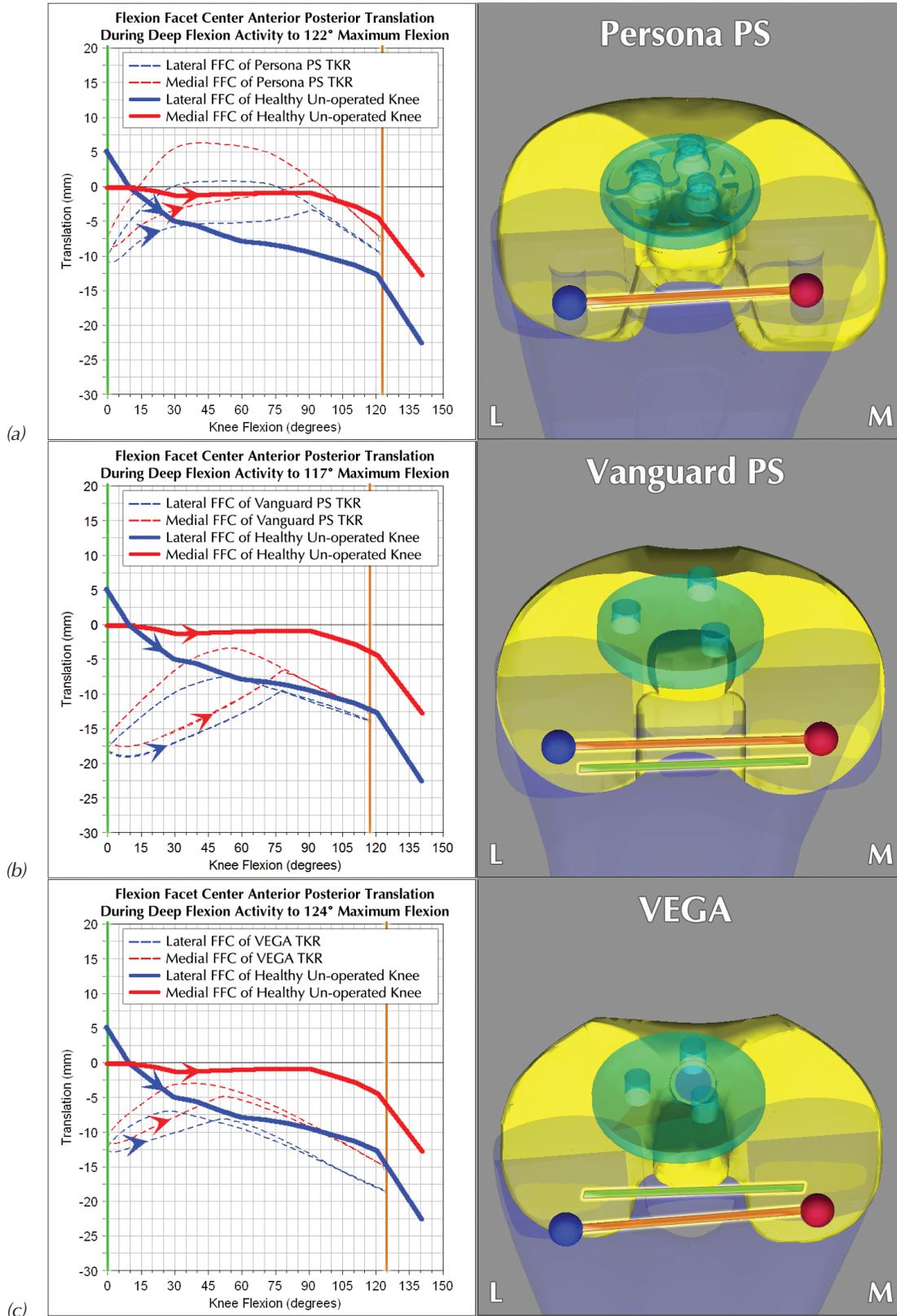


Figure 5

## DISCUSSION

The dashed blue and red lines in the plots on the left of Figures 4 and 5 indicate the anterior/posterior translation of the FFCs for each design. In general the FFCs follow a counter clockwise path on the plot, taking a more posteriorly located path (indicated with arrow heads) as flexion increases and a more anterior path as the high flexion activity returns to full extension. A smaller sized loop indicates that the design component geometries tightly control anterior posterior translation. All designs studied slide anteriorly in various amounts while flexion increases, a motion paradoxical to the healthy intact knee. The direction is reversed when the femoral cam engages the tibial post, forcing movement in the posterior direction while flexion continues to its maximum.

Close inspection of the component motions on the right reveal that contact areas (light yellow patches) are often coincident with the FFC marker from a superior view, but can readily diverge when anterior or posterior forces applied to the femoral component cause the contact area to traverse much farther than the component itself translates. This illustrates why the extent of the burnishing and abrasive wear scars found in tibial insert retrievals are much greater than the motion that the femoral component itself can achieve. FFCs do not indicate centroid locations of contact area, but rather serve as reference points to help visualize motion of the femoral component relative to the tibia.

The Legacy LPS-Flex Fixed achieved the highest flexion angle of  $144^\circ$  among the designs studied. Design features contributing to this outcome are illustrated from a medial sagittal view (Figure 6). The post/cam mechanism promotes femoral component contact near the posterior edge of the tibial insert (contact area outlined in red), and additional posterior femoral bone resection allow deeper flexion to be achieved before bony impingement occurs. A posterior sloped proximal tibia bone resection of  $7^\circ$  further contributes to high flexion, however at the observed expense of impingement of the anterior aspect of the femoral cam and tibial post at  $4^\circ$  of flexion.

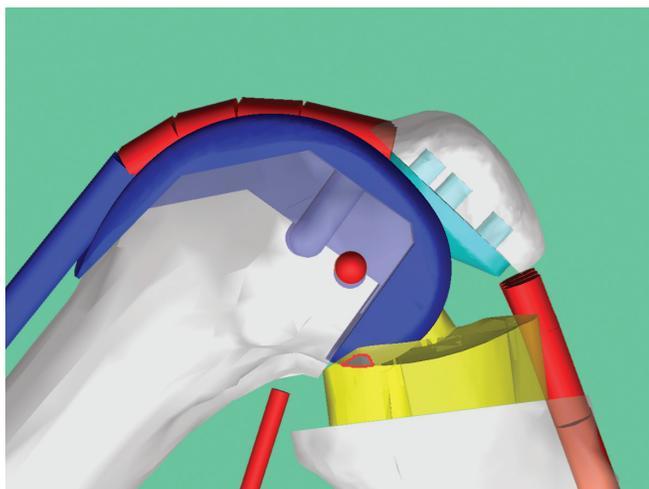


Figure 6: Medial view of Legacy LPS-Flex Fixed at its maximum flexion of  $144^\circ$ .

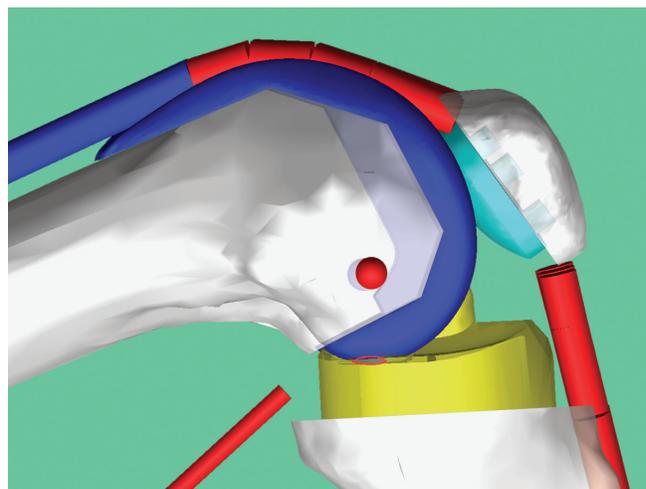


Figure 7: Medial view of Vanguard PS at its maximum flexion of  $117^\circ$ .

The medial sagittal view of the Vanguard PS (Figure 7) achieved the lowest flexion angle of  $117^\circ$  among the designs studied. This result illustrates a design trade off of very high flexion for bone stock preservation and tibial insert longevity. The Vanguard PS is a more conservative bone preserving design with a less aggressive posterior femoral and  $3^\circ$  proximal tibia bone resection than the Legacy LPS-Flex Fixed. The post/cam mechanism promotes femoral component contact further away from the posterior edge of the tibial insert (contact area outlined in red) allowing more polymer support during demanding high flexion activities.

The method used in this study of posterior femoral bone cut surface impingement with the tibial insert defining maximum flexion is controversial. Although posterior femoral bone cut/tibial insert impingement is a clinical reality evidenced by component retrieval studies<sup>9</sup>, many patients achieve higher flexion.

## CONCLUSIONS

None of the posterior stabilized total knee arthroplasty designs investigated in this study were able to closely replicate the motion of the healthy intact knee during a high flexion activity, which remains an elusive goal of contemporary TKA design. The Journey II, and to a lesser degree the VEGA, designs were able to achieve early femoral rollback and external femoral rotation, both hallmarks of healthy intact knee motion. Most designs displayed femoral paradoxical motion of 5 to 10 millimeters of anterior sliding before post/cam engagement and exhibited lateral or no pivoting.

All of the designs investigated in this study achieved high flexion by western patient standards, with maximum flexion angles ranging between 117° and 144°. The results of this study suggest that higher flexion can be achieved at the expense of additional loss of bone stock with aggressive posterior femoral and proximal tibial resection. The consequences of this approach may include premature fixation failures<sup>2</sup> and increased tibial insert damage<sup>10</sup>.

The value of this study lies in its ability to hold surgical and patient variables constant, allowing focus on the effect of TKA design on knee motion. Dynamic, validated computational models expand the methodologies available to investigate and better understand factors influencing knee kinematics following total knee arthroplasty. This holds great promise for further total knee design optimization and improvements in surgical procedure leading to better patient outcomes.

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