Chapter 6
Stability Characteristics of the Tibial-Femoral and Patellar-Femoral Articulations*

A.S. Greenwald

1 Introduction

The Low Contact Stress (LCS) knee system was introduced in 1977 as an attempt to improve the survivorship and clinical function of total knee replacement. From its outset, the design goals of this mobile bearing knee system were to decrease the prospect of polyethylene material damage, minimize constraint and optimize patient implant kinematics. The gait-congruent LCS design, in its origin, was proposed as a system inclusive of both cruciate-sparing meniscal bearing and PCL-sacrificing rotating platform variants, with the latter gaining the majority of popular usage over time. Both systems employ a dynamic tracking metal-backed rotating patellar component that articulates with the femoral sulcus.

Fixed plateau knee designs have two extremes. At one end, minimal geometric constraint against displacement results in small contact areas and high contact stresses. The benefit is the transmission of minimal torques to fixation interfaces but at the increased risk of polyethylene articulation damage. Alternately, more highly conforming devices distribute peak stresses over increased contact surfaces, but with the consequence of increased constraint and interface force transmission.

Mobile bearing knees represent a design optimization of these extremes. They offer significant increases in articulation conformity, reducing contact stresses, while at the same time, through insert mobility, they minimize constraint forces transferred to fixation interfaces [1, 3, 6, 9, 13].

Dual surface articulation between a UHMWPE insert and a metallic femoral and tibial tray characterizes this unique knee design. For this reason, the Food and Drug Administration required both laboratory and prospective clinical evaluations to demonstrate its safety and efficacy [2, 4, 7, 11, 18]. This paper describes the stability characteristics of the LCS meniscal bearing and rotating platform designs.

2 Tibial-Femoral Stability

The restoration of normal knee function through surgical reconstruction is highly dependent upon load sharing between the implant, surrounding ligaments and other supporting soft tissue structures. Excision, surgical release and progressive pathological weakening of ligamentous structures results in an increased dependency upon the implant system for stability. Intrinsically stable, achieved in non-hinged, total knee replacements through geometric variation of the condylar surfaces, is influenced by the relationship to the active and passive soft tissue structures. Stability in this regard is the capacity of the implant to limit rotational, anterior-posterior, and medial-lateral displacements to within normal physiologic ranges [8, 9, 10].

The bar graphs depicted in Fig. 1 and 2 describe the stability characteristics of LCS system designs under physiological compression loading for both posterior and rotational directions. Comparative data for fixed plateau designs has also been included [16]. The horizontal lines depict the maximum posterior shear force and rotatory torque that have been reported for normal knee function [15, 17]. One may deduce that implants whose bars are below the reference line are indicative of the need for soft tissue involvement to achieve stability under the forces of gait. Parenthetically, it may be appreciated that the distance between the top of the bar graph and the reference line is a measure of the soft tissue competency demanded to achieve overall stability. This competency may be provided by active collateral, cruciate and capsular soft tissue structures as well as by particular muscle activity. It is of further interest that an early argument against meniscal bearing use was that the bearing tracks would become soft tissue ingrown, inhibiting rotatory motion. A further experiment in

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which the bearings were fixed indicates a significant increase in the intrinsic rotational stability provided by the geometry of the condyles. This points to a significantly more constrained system, less dependent on soft tissue involvement and potentially transferring higher interface stresses to the implant-bone interface. Given the low incidence of in vivo tibial component loosening with the LCS designs, it is an inference of bearing motion [2, 11]. Like many of their fixed plateau counterparts, the LCS is seen to be dependent on soft tissue involvement for stability during function. This load sharing enhanced by plateau motion is regarded as a positive factor in achieving in vivo implant durability.

3 Patellar-Femoral Stability

While the need for patellar resurfacing remains a contentious issue, the majority of American use favors replacement. Yet patellar replacement remains one of the more frequent reasons for knee failure. Clinical outcome is dependent upon patient factors, technical proficiency and implant design. The normal patella tracks in an anatomical groove on the anterior distal femur. Compressive and lateral forces acting at the patellar-femoral joint increase with knee flexion. In the normal valgus knee, a lateral resultant force is produced by the quadriceps mechanism. The magnitude of this reaction depends upon individual anatomy and soft tissue balance. Although not precisely defined, estimates suggest significant force generation. Excessive lateral forces can cause patellar subluxation or dislocation contributing to component failure.

A series of experiments to define the resistance of contemporary patellar-femoral articulations to lateral subluxation under physiological compressive loads inclusive of the LCS lend insight into anticipated design performance. The bar graphs in Fig. 3–5 depict the intrinsic lateral stability of the LCS patellar-femoral implant at varying degrees of flexion. For comparative purposes, data for all-polyethylene, cemented patellar designs are included [19]. The estimated lateral load components for the normal knee are represented by the horizontal line [5, 12, 14]. The interaction of condylar and patellar geometry defines the resistance offered to lateral subluxation. The lateral stabilities of the LCS system, as well as those of the other designs, vary with knee flexion angle. Inherent geometrical constraint should correspond with anticipated in vivo lateral loading. The results suggest that sufficient intrinsic stability is provided by the patellar-femoral geometry of the LCS design over the gait cycle. Overall patellar-femoral stability is defined by the interaction between these geometries and the ex-