ANALYSIS OF IN VIVO CROSSING MOTION IN TOTAL KNEE REPLACEMENTS

W. Gregory Sawyer (1), Matthew A. Hamilton (1), Matthew C. Sucec (1), Benjamin J. Fregly (1,2,3), and Scott A. Banks (4)

(1) Dept. of Mechanical & Aerospace Engineering
University of Florida
Gainesville, FL

(2) Department of Biomedical Engineering
University of Florida
Gainesville, FL

(3) Department of Orthopaedics & Rehabilitation
University of Florida
Gainesville, FL

(4) Orthopaedic Research Lab
The Biomotion Foundation
West Palm Beach, FL

INTRODUCTION

Debris particle generation of ultra-high-molecular-weight polyethylene (UHMWPE) remains a clinical issue in total joint replacements such as knees and hips. Significant advances in modeling and understanding the origin of wear debris liberation in total hip replacements have been made over the past decade. One recently-discovered aspect is the importance of sliding direction on wear in UHMWPE [1,2]. Orders of magnitude changes in wear rate with increasing degree of crossing motion have been reported.

In contrast to the hip, which is axi-symmetric and conformal, the knee produces complex motions that have prevented detailed study of tibial insert crossing patterns. Locus plots (Fig. 1a), which seek to represent the kinematics by following the most probable trajectory of contact on the surface, do not capture the relative motions experienced by particular surface locations (Fig. 1a). While slip velocity vectors can be plotted for individual surface locations (Fig. 1b) [2], they cannot be plotted for all locations simultaneously to visualize crossing on the entire surface. Thus, a new approach is needed to visualize the extent of crossing experienced by *all* locations on the surface *simultaneously* over an entire activity cycle. This study presents such an approach and evaluates it using *in vivo* patient-specific kinematics.

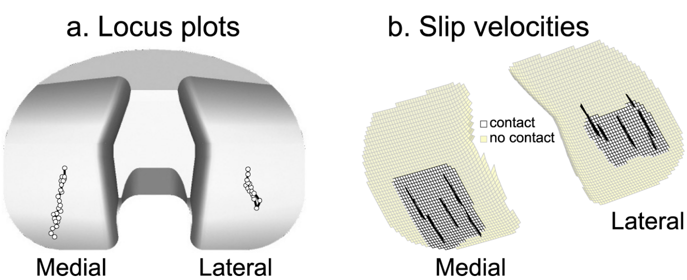

Figure 1. (a) Locus plot and (b) slip velocity representations of crossing motions on the tibial insert contact surfaces based on in vivo stair kinematic data.

METHODS

Kinematic data previously collected from one total knee arthroplasty patient (female, age 65 at surgery, height 170 cm, mass 70 kg) were used in this study [3]. The patient received a cemented posterior cruciate ligament retaining prosthesis (Series 7000, Stryker Howmedica Osteonics, Allendale, NJ) with a 6.8 mm thick tibial insert. The patient performed treadmill gait and stair rise/descent activities during fluoroscopic motion analysis [4] and gave written informed consent to participate [3]. Kinematic data from one representative cycle of each activity were averaged in 5° increments of knee flexion for stair and 1% increments for gait including stance and swing phases. Cycle duration was normalized to 1 sec for both activities.

Dynamic simulations to predict *in vivo* tibial insert contact pressures and slip velocities were created by incorporating an elastic contact model into the commercial multibody dynamics code Pro/MECHANICA MOTION (Parametric Technology, Waltham, MA). The contact model treats the tibial insert as an elastic foundation [5,6] contacting a rigid femoral component, where contact pressures are calculated on a grid of mutually-independent elements covering the insert surfaces [5]. For any element, given the interpenetration δ between the undeformed surfaces, the contact pressure p acting on the element can be calculated from [5,6]

$$p = \frac{(1-\nu)E}{((1+\nu)((1-2\nu)h}\delta \qquad (1)$$

where E is Young's modulus of the elastic layer, ν is Poisson's ratio of the elastic layer, and h is the layer thickness. The resulting element pressures are replaced with a single equivalent force and torque applied to both bodies for purposes of multibody dynamic simulation.

The simulations were driven with a combination of the *in vivo* fluoroscopic data and assumed loading conditions. Three DOFs (anterior-posterior translation, internal-external rotation, and flexion) were prescribed to match fluoroscopically measured gait and stair kinematics. The three remaining DOFs were numerically integrated to predict their motion. An axial force was applied vertically downward to the femoral component to produce a 70% medial-30% lateral load split at 0º flexion [7]. The force magnitude for each activity was defined by scaling a vertical ground reaction force curve from a patient of similar age, height, weight, and knee flexion characteristics to be between 0.25 and 3.0 BW [8].

The predicted contact pressures and slip velocities for individual surface elements were used as inputs to the proposed analysis of crossing motion. The analysis observes the motion of the femoral component relative to each element on the tibial insert surface and is based on sequential calculation of three quantities. The first is tribological intensity τ defined as

$$\tau = p\left|\mathbf{d}\right| \tag{2}$$

where $\mathbf{d}$ is the slip vector created by multiplying the instantaneous slip velocity with the simulation time increment. Equation (2) is related to Archard's classic wear law and represents the fact that crossing is detrimental only if the element is in contact and experiencing relative motion. The second quantity is dominant orientation of tribological intensity θ^* defined as

$$\theta^* = \sum_{i=1}^{n} \tau_i \theta_i \Bigg/ \sum_{i=1}^{n} \tau_i \tag{3}$$

where i indicates the current time during the activity and θ_i indicates the instantaneous crossing orientation relative to a fixed medial-lateral axis. θ_i is restricted to be between 0 and π since reciprocating motion would produce a single polymer orientation. Equation (3) weights each θ_i by the corresponding τ_i and then normalizes to define most probable direction of polymer orientation for a particular element. The last quantity is crossing intensity σ defined as

$$\sigma = \sqrt{\sum_{i=1}^{n} (\tau_i \Delta\theta_i)^2 \Bigg/ n} \tag{4}$$

where $\Delta\theta_i = \theta^* - \theta_i$. Equation (4) represents the spread of tribological intensity about the dominant orientation direction θ^*. The best-case scenario, uni-directional motion in the direction of θ^*, would produce $\sigma = 0$ since no crossing would occur.

To turn σ into a practical measure, the worst-case crossing scenario σ_o must also be defined. This occurs for uniform circular motion with constant pressure, or when $\theta^* = \pi/2$. Normalized crossing intensity σ^* defined as

$$\sigma^* = \sigma / \sigma_o \tag{5}$$

can then be used as a single dimensionless measure of crossing intensity. In this study, σ^* was calculated for every element on the medial and lateral tibial insert contact surfaces for both the gait and stair simulations.

RESULTS

The patient showed limited crossing motion with the maximum value of σ^* being about 0.04 for gait and 0.09 for stair (Fig. 2). These values correspond to bi-directional counterface motions of at most 10º. For both activities, the greatest normalized crossing intensities occurred on the lateral side, consistent with the lateral pivoting pattern observed in the patient's kinematic data and tibial insert retrieved post-mortem. Though the corresponding tribological intensity plots (not shown) closely resembled Fig. 2, the locations of maximum τ did not always match to the locations of maximum σ^*. Furthermore, the same value of τ in the medial and lateral compartments produced very different values of σ^*.

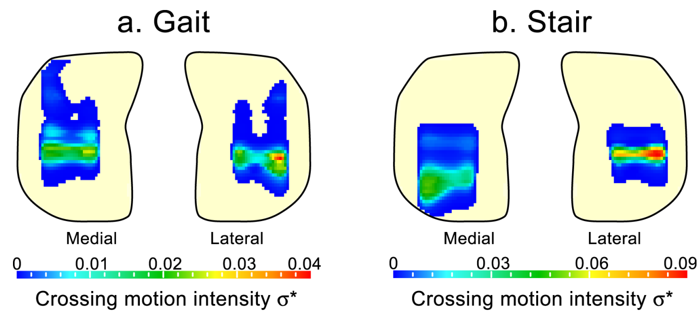

Figure 2. Normalized crossing motion intensities for (a) gait and (b) stair activities based on in vivo kinematic data.

DISCUSSION

This study proposes a new method for calculating the intensity of crossing motions in total joint replacements. The method overcomes the limitations of previous methods by permitting visualization of overall crossing intensity for *all* elements on the surface *simultaneously* for an entire activity cycle. The crossing motions observed for this patient suggest that uniform bi-directional patterns with 10° of included angle are a reasonable screening motion for pin-on-disk testing. It is unclear how significant this degree of crossing is to the tribological behavior of the UHMWPE implanted in knees.

Conforming knee prostheses designed to pivot on the medial or lateral side are currently in use. Evaluation of the potential tribological impact of such designs requires evaluation of both tribological intensity and crossing intensity. Perhaps, overall tribological severity is the product of the two.

REFERENCES

1. Bragdon, C.R., O'Conner, D.O., Lowenstein, J.D., Jasty, M., and Syniuta, W.D., 1996, "The importance of multidirectional motion on the wear of polyethylene," IMechE Part H: Journal of Engineering in Medicine, Vol. 210, pp. 157-165.
2. Wang, A., 2001, "A unified theory of wear for ultra-high molecular weight polyethylene in multi-directional sliding," Wear, Vol. 248, pp. 38-47.
3. Harman M.K., Banks S.A., and Hodge W.A:, 2001, "Polyethylene damage and knee kinematics after total knee arthroplasty," Clinical Orthopaedics and Related Research, 392, pp. 383-393.
4. Banks, S.A. and Hodge, W.A., 1996, "Accurate measurement of three-dimensional knee replacement kinematics using single-plane fluoroscopy," IEEE Transactions on Biomedical Engineering, 43, pp. 638-649.
5. Johnson, K.L., 1985, *Contact Mechanics*, Cambridge University Press, Cambridge.
6. An, K.N., Himenso, S., Tsumura, H., Kawai, T., and Chao, E.Y.S., 1990, "Pressure distribution on articular surfaces: application to joint stability analysis," Journal of Biomechanics, 23, pp. 1013-1020.
7. Schipplein, O.D. and Andriacchi, T.P., 1991, "Interaction between active and passive knee stabilizers during level walking," Journal of Orthopaedic Research, 9, pp. 113-119.
8. Taylor, S.J.G. and Walker, P.S., 2001, "Force and moments telemetered from two distal femoral replacements during various activities," Journal of Biomechanics, 34, pp. 839-848.