COMPUTATIONAL DETERMINATION OF IN VIVO MEDIAL AND LATERAL TIBIAL FORCES DURING GAIT

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INTRODUCTION

Asymmetric loading between the medial and lateral compartments of the knee has been hypothesized to contribute to the development of knee osteoarthritis (OA) [1]. In artificial knees, asymmetric forces exerted on the tibial insert may contribute to mechanical failure as well as loosening of the implant [2]. Studies using statically determinate muscle models, video-based motion analysis, and external force measurements have predicted larger contact forces on the medial than on the lateral side of the knee during gait. Furthermore, the predicted medial-lateral load split showed considerable variation over the course of the gait cycle [3]. However, the actual *in vivo* distribution of tibial contact forces during gait remains unknown.

Recently, the first instrumented knee implant capable of measuring the *in vivo* axial loads applied to the tibia has been reported [4]. This device allows the total axial load and center of pressure (CoP) to be calculated under dynamic conditions from force measurements made by four uniaxial load cells located at the corners of the tibial tray. By augmenting these *in vivo* load measurements with *in vivo* motion measurements performed simultaneously using video fluoroscopy [5], one can use a multibody dynamic contact model to calculate the time history of medial and lateral contact forces during dynamic weight-bearing activities such as gait.

This study uses the instrumented knee implant described above, along with fluoroscopic motion analysis and a dynamic contact model, to determine the medial-lateral contact force distribution on the tibia *in vivo* during one cycle of gait. The reliability of the calculated medial and lateral contact forces is evaluated based on the contact model's ability to reproduce the experimentally measured medial-lateral (ML) and anterior-posterior (AP) CoP locations.

METHODS

Data were collected from one total knee arthroplasty patient (male, right knee, age 80, mass 67 kg) eight months after surgery. Institutional review board approval and patient informed consent were obtained. The patient received a custom tibial prosthesis design instrumented with force transducers, a microtransmitter, and an antenna. *In vivo* tibial forces were recorded simultaneously with either fluoroscopic motion analysis data (treadmill gait) or video-based motion analysis and ground reaction force data (overground gait). Only treadmill gait was studied in this paper, and the gait cycle was defined to start at right heel strike. This event on the treadmill was identified by determining the location in the instrumented knee load data where the vertical ground reaction force measured during overground gait become nonzero.

A dynamic contact model of the patient's knee implant was constructed to predict *in vivo* contact forces, pressures, and areas on medial and lateral contact surfaces of the tibial insert. The model was implemented within the Pro/MECHANICA MOTION simulation environment (PTC, Waltham, MA) (Fig. 1) and used a previously reported elastic foundation deformable contact model with linear material properties [6]. A 6 degree-of-freedom (DOF) joint between the fixed femoral component and moving tibial insert was used to measure relative (i.e., joint) kinematics for contact calculations. Femoral AP translation, internal-external rotation, and flexion-extension were prescribed to match the fluoroscopically measured kinematics while the other three DOFs were predicted via forward dynamic simulation. The location at which the axial force was applied to the tibial tray was prescribed to match the CoP measured experimentally.

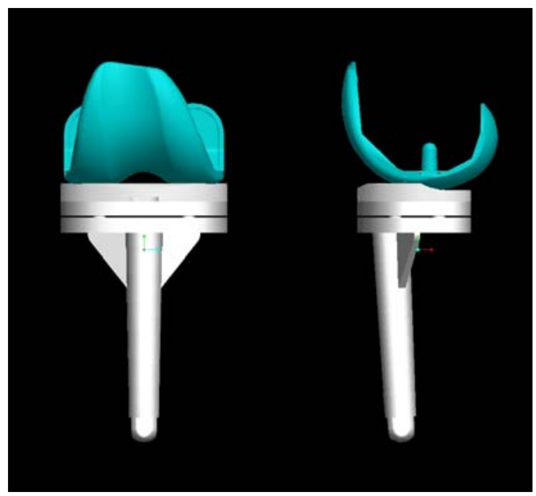

Figure 1: Dynamic contact model of the instrumented knee implant developed within the Pro/MECHANICA MOTION simulation environment.

The medial and lateral contact forces acting on the tibial insert were calculated from the contact pressures acting across the surfaces. The CoP location in the ML and AP directions was also calculated from the model for comparison with the experimentally measured CoP locations.

RESULTS

The dynamic contact model was able to match the experimentally measured CoP over most of the gait cycle. The CoP calculated from the instrumented knee implant remained near the geometric center of the tibial insert (Fig. 2). During stance phase (red dots), the CoP shifted slightly to the medial and lateral sides, while during swing phase (green dots), it remained on the lateral side. Without changing any

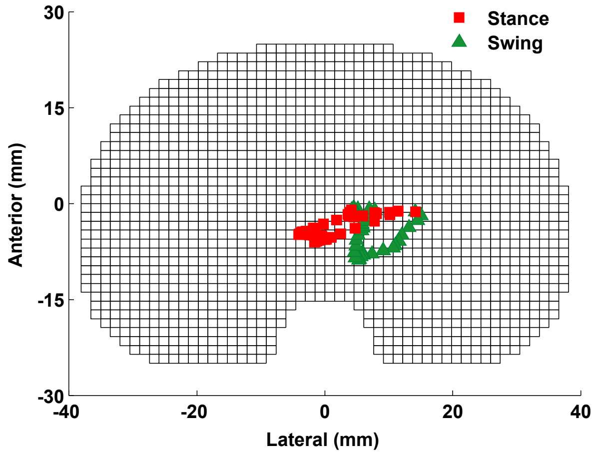

Figure 2: Visualization of center of pressure calculated from the instrumented knee implant during treadmill gait.

fluoroscopically measured motion inputs, the dynamic contact simulation reproduced the ML location of the CoP extremely closely throughout the gait cycle with a root-mean-square (RMS) error of 0.4 mm (Fig. 3a). In contrast, it matched the AP location of the CoP well only during stance phase, with a RMS error of 4.7 mm over the entire cycle. When the fluoroscopically measured AP translation was adjusted within its range of experimental error (± 1 mm), the RMS error in the AP direction was reduced to 3.3 mm over the entire cycle and 0.7 mm during stance phase (Fig. 3b), with the largest errors still occurring during swing phase.

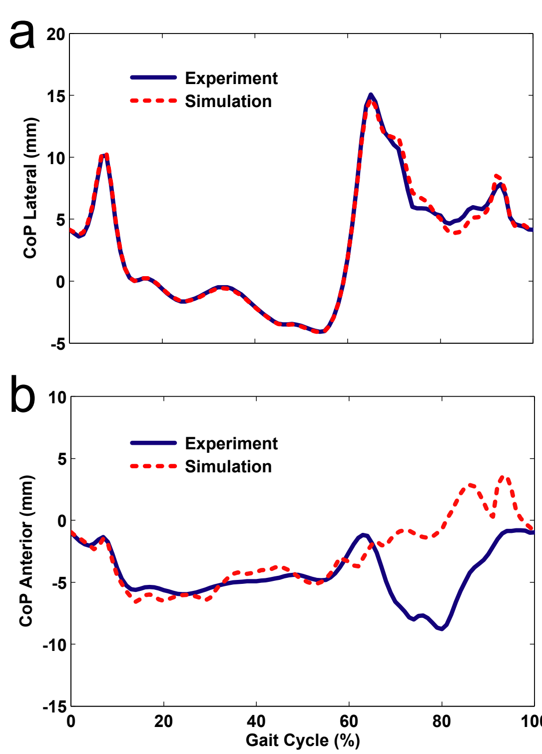

Figure 3: Comparison of experimental and simulated CoP location in a) the ML direction (lateral is positive), and b) the AP direction (anterior is positive).

When the total axial load measured experimentally (Fig. 4, blue curve) was decomposed into medial (green curve) and lateral (red curve) loads by the contact model, the ratio of medial to total force ranged from 17.1% (toe off, 65% of gait cycle) to 59.3% (late stance phase, 55% of gait cycle). During mid-stance when the largest forces were applied, more than 50% of the load passed through the medial compartment.

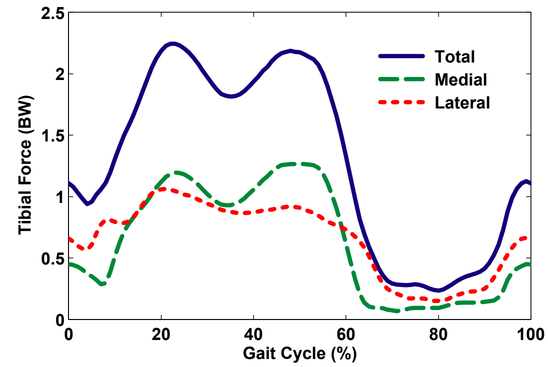

Figure 4: Medial and lateral contact forces calculated by the dynamic contact model during gait**.**

DISCUSSION

The approximately equal distribution of medial and lateral forces during mid-stance challenges the belief that the majority of the load passes through the medial compartment during gait. Medial knee OA may be more common than lateral knee OA because of the greater rate of loading in the medial compartment. This higher loading rate could lead to higher fatigue damage of the cartilage bearing surface.

Given that the CoP locations measured by the instrumented implant are accurate to within ± 2 mm [4], the errors in the predicted CoP location were small apart from the AP direction during swing phase. The experimental CoP calculation involves dividing by the total axial load, and the error in predicted AP CoP location was highly correlated ($R^2 = 0.88$) with the inverse of the applied load. Thus, sensitivity of the experimental CoP calculations to errors in the measured axial load may help explain the larger CoP prediction errors during swing phase.

CONCLUSIONS

In summary, this study used a novel combination *in vivo* load and motion measurements with a computational model to predict the medial-lateral force distribution in an implanted knee during gait. The results indicate that in well aligned implanted knees, the load split during mid stance phase is more equal than previously predicted.

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ACKNOWLEDGEMENTS

This work was supported by a NSF CAREER award to B.J.F. and by the Biomotion Foundation of Palm Beach, FL.