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Patient-Specific Computer Model of Dynamic Squatting After Total Knee Arthroplasty

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ABSTRACT

Knee forces are highly relevant to performance after total knee arthroplasty especially during high flexion activities such as squatting. We constructed subject-specific models of two patients implanted with instrumented knee prostheses that measured knee forces in vivo. In vivo peak forces ranged from 2.2 to 2.3 times bodyweight but peaked at different flexion angles based on the type of squatting activity. Our model predicted tibiofemoral contact force with reasonable accuracy in both subjects. This model can be a very useful tool to predict the effect of surgical techniques and component alignment on contact forces. In addition, this model could be used for implant design development, to enhance knee function, to predict forces generated during other activities, and for predicting clinical outcomes.

Total knee arthroplasty (TKA) has become one of the most successful orthopedic procedures in providing pain relief and improving knee function, with reported survival rates of greater than 90% after 15 years [1,2]. The success of TKA is dependent on many factors including preoperative status, surgical technique, and the design and materials of the components. While survival rates are high, functional outcomes that facilitate common activities involving deep knee flexion such as kneeling, squatting, and sitting cross-legged are rarely achieved [3,4]. Knee contact force during activities after TKA is very important since it affects component wear and implant loosening. Knee contact force is related directly to the transmission of stresses through the implant, which include contact stresses generated at the bearing surface and subsurface, stresses at the implant–cement–bone interface, and stresses transmitted to underlying bone [5].

Previous studies have measured knee contact forces in cadaver models and biomechanical simulators. However, there are technical challenges in applying high physiologic loads to the knee joint coupled with the inherent weaknesses of extrapolating in vitro results to vivo function. While several computational models have predicted knee contact force, these reports vary widely based on the modeling approach and the assumptions inherent to the model. Predictions of tibiofemoral forces made by computer models have also varied widely for the same activity [6–9]. For example, peak forces predicted for walking prior to the availability of in vivo data ranged from 1.8 × BW to 8.1 × BW (times of bodyweight, reviewed by [10]). Peak forces predicted by computer models for squatting have also been variable, ranging from 3.4 × BW to 7.3 × BW [11–13].

The complexity of modeling the knee is in part due to tricompartmental contact with joint stability governed primarily by soft tissues. Knee contact forces can vary widely among patients due to differences in subject anatomy, bodyweight, and kinematic patterns. For accuracy, clinically relevant predictions and a subject-specific approach may be necessary to account for this interpatient variability. Subject-specific approaches have been reported with some validation of knee forces predicted during walking [10]. However, during weight-bearing deep knee bend activities such as squatting, the knee forces of the computer-generated model were much higher and until very recently these predictions had not yet been validated in vivo [11–14]. One recent study validated a subject-specific model of squatting with in vivo measured experimental knee forces and EMG [15]. The purpose of this study was to extend that approach by constructing subject-specific computer models of two different patients implanted with instrumented knee arthroplasty components, each performing two variations of a dynamic closed-kinetic-chain squatting activity. Patient kinematics and ground reaction forces were input into the model and the predicted tibiofemoral contact forces were compared to in vivo measured forces.

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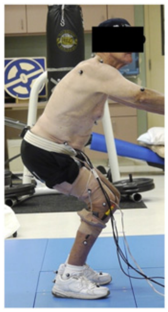

Fig. 1. Photograph of patient performing squatting motion.

Methods

Patient Information

Institutional review board approval and informed patient consents were obtained for this study. Three patients had been implanted with a custom tibial prosthesis instrumented with force transducers and a telemetry system [5,10,16–23]. The tibial prosthesis was customized to house force sensors and a telemetry system. The sensors measured three components of force and three components of moment acting on the tibial tray. These six measurements can be used to calculate the contact forces in the medial and lateral compartment. Two male patients were selected for this study (83-year-old, 69.5 kg, right knee; 88-year-old, 76.3 kg, left knee). One patient had no significant arthritis in his contralateral knee. The other patient had the contralateral knee replaced with the same design but without the custom tibia with electronics. Details of the implant design and surgical technique have been previously reported [18,24]. The distal femoral cut was made at a nominal 6° valgus to the anatomic axis of the femur using intramedullary alignment, while the posterior femoral cut was made in 3° external rotation with reference to the posterior condyles. The tibial bone cut was made at a nominal 90° to the long axis without any posterior slope. Standard cruciate-retaining Natural Knee® II (Zimmer) femoral components were cemented. The custom instrumented tibial prosthesis was cemented, and a 10 mm thick cruciate-retaining polyethylene insert was implanted. Measurements were made on postoperative computerized tomographic scans to obtain subject-specific femoral and tibial component alignment.

Experimental Measurement of Knee Forces In Vivo

The patients were three years postoperative at the time this study was conducted. Two different squatting activities were performed with both feet parallel to each other. For the first squatting activity patients were instructed to squat to the maximum knee flexion angle within tolerance and with the trunk flexed to a patient-preferred degree. For the second squatting activity, the patients were instructed to keep their hands on their hips and their upper body as upright as possible within tolerance (to minimize trunk flexion and maximize knee flexion moment). These variations were chosen to alter the external flexion moment on the knee. Each squatting cycle was repeated three times. Skin marker-based video motion analysis was used to record knee kinematics, and axial ground reaction forces were measured under each foot (Fig. 1).

Patient-specific Computer Model

Preoperative and postoperative computer tomographic scans were reconstructed to extract tibiofemoral bone geometry using MIMICS (Materialise, Leuven, Belgium). Computer-aided design models of the components were directly aligned to the 3-dimensional bone models as we have previously reported (Fig. 2) [25]. Bone and implant geometry was imported into a dynamic, musculoskeletal modeling program (LifeMOD/BodySIM 2008, LifeModeler, Inc., San Clemente, California). We previously validated a LifeMOD/KneeSim model using cadaver data measured in an Oxford knee rig [26]. A 14-segment model (head, trunk, upper arms, forearms, hands, thighs, legs, and feet) was initially constructed based on published generic anthropometric data (age, gender, bodyweight, and height) [27]. The body segments were then scaled, using measurements obtained from each patient. Each lower limb was then populated with 17 base muscles (gluteus maximus [2], gluteus medius [2], psoas major, iliopsoas, rectus femoris, vastus medialis, vastus lateralis, adductor magnus, biceps femoris [2], semitendinosus, tibialis anterior, soleus, and gastrocnemius [2]), three ligaments (medial collateral ligament [MCL], lateral collateral ligament [LCL], and posterior cruciate ligament [PCL]) and two tendons (patellar and quadriceps). Knee ligament attachments were based on subject-specific anatomic landmarks obtained from the preoperative CT scans, and ligaments were modeled as nonlinear springs using previously published material properties [28]. The quadriceps and patellar tendons were modeled with contact-based rigid bodies to simulate contact and wrapping around the trochlear groove and tibial insert, respectively (Fig. 3).

Computer Simulation of Squatting

The computer-generated image of the simulation and photograph for one of the subjects during the squatting activity is shown in Fig. 4. The simulation was carried out in two steps and has been previously described in detail [15]. In the first step, skin marker-based motion data were used to prescribe the trunk, hip, knee, and ankle kinematics for an inverse-dynamics computation. During this step, joint torques and changes in muscle lengths were recorded throughout the activity cycle. In the next step, forward dynamics was used to compute the

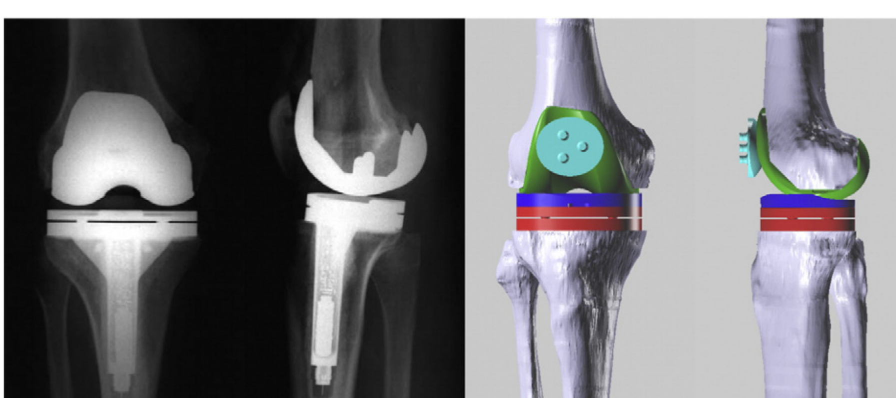

Fig. 2. Radiographs and corresponding CT-generated models of the implanted knee.