VALIDATION OF A COMPUTATIONAL MODEL USED TO CHARACTERIZE THE PATELLOFEMORAL CONTACT PRESSURE DISTRIBUTION

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INTRODUCTION
Patellofemoral pain and arthrosis are often attributed to malalignment. The quadriceps and patella tendon forces acting on the patella have a natural lateral orientation, which is characterized by the Q-angle (the angle formed between the patella tendon and the primary orientation of the quadriceps). When the Q-angle is large, the lateral force applied to the patella is believed to overload the lateral cartilage. Several surgical procedures focus on reducing the Q-angle and/or the total force applied to the patella by altering the orientation of the patella tendon or the primary quadriceps orientation. How these procedures influence the patellofemoral contact pressure distribution is not well understood, however, and anatomical variations between patients may drastically influence the effectiveness of each procedure. A computational model has been created to help orthopaedic surgeons predict how natural and surgically created variations in patellofemoral anatomy and loading conditions influence the pressure applied to the cartilage during functional activities [1]. The current study was performed to determine how well the model predicts variations in the patellofemoral force and pressure distributions.

METHODS
Four cadaver knees were used to experimentally measure the influence of the Q-angle on the patellofemoral contact pressure distribution. The knees were flexed on a testing rig that simulated a 50% bodyweight during squats. An 87 N vertical load applied at the hip induced flexion, which was resisted by a load applied through the quadriceps tendon. Each knee was tested with the quadriceps loading mechanism aligned with the femur (average initial Q-angle = 6°), and after reorienting the quadriceps to simulate a 5° Q-angle decrease and increase. A pressure sensitive sensor (I-scan, Tekscan) was glued to the articular surface of each patella, and reflective markers were secured to the tibia and patella to track the motion of each bone (PCReflex, Qualisys). Anatomical landmarks of the patella were digitized on the film, and the force and pressure distributions were mapped onto a two-dimensional patella surface throughout flexion [2].

A geometric model of each knee was reconstructed from CT data. Each tibia was flexed in 10° increments to a maximum of 70° about the femur, which was the maximum flexion angle achieved by all knees experimentally. The patella was flexed based on the kinematic data, and anatomically positioned by aligning the posterior patella surface in the trochlear groove. No shift, rotation or tilt was applied to the patella. At each flexion angle, a static analysis of the forces acting on the tibia was performed to calculate the patella tendon force, which in turn used to quantify the quadriceps force [3]. To calculate the contact pressure distribution, a surface was created midway between the patella and femur. Approximately 3000 compressive springs were placed on the surface to model the cartilage, which was assigned an elastic modulus of 5 MPa. For all three Q-angles, the resultant force and moment acting at the centroid of the patella were calculated, and the patellofemoral contact pressure distribution was quantified by minimizing the total potential energy within the system of springs. The percentage of the patellofemoral force applied to the lateral cartilage and the peak medial and lateral contact pressures were quantified to compare the computational and the experimental results. For the computational and the experimental results, the peak contact pressure values for each knee were normalized by the peak lateral pressure at 70° of flexion.

RESULTS
The force distribution trends were similar for the experimental results and the computational results. For the normal Q-angle at 70° of flexion, the percentage of force applied to the lateral cartilage was 55% ± 5% (average ± standard deviation) for the computational model, compared to 56% ± 9% experimentally. The lateral force percentage was less than 50% for only one knee, with a computational value of 48% and an experimental value of 42%. Increasing the Q-angle increased the percentage of force applied to the lateral cartilage by 7% ± 2% computationally and 2% ± 2% experimentally, while decreasing the Q-angle decreased the percentage of lateral force by 6% ± 2% computationally and 3% ± 2% experimentally.

The computational model accurately showed that the maximum lateral pressure increased with the flexion angle and the Q-angle (Fig. 1). For the normal Q-angle, the maximum medial and lateral pressures were similar in magnitude. Both computationally and experimentally, decreasing the Q-angle tended to increase the maximum medial pressure. Both the computational and experimental results also showed that increasing the Q-angle tended to shift the region of peak pressure laterally on the patella (Fig. 2).

DISCUSSION
The results indicate that the computational model can predict the influence of a variation in the Q-angle on the distribution of force and pressure within the patellofemoral joint with enough accuracy to be clinically useful. No other patellofemoral model that the authors are aware of has been directly compared to experimental measurements of patellofemoral pressure. This type of computational analysis could be used to characterize how surgical procedures performed to alter the forces applied to the patella influence the cartilage, providing data to evaluate the effectiveness of each procedure. By incorporating the anatomy of individual patients into the model, the computational analysis could also help surgeons optimize surgical procedures for their patients based on the force applied to the patellofemoral cartilage.

REFERENCES

LISTING FOR ADDITIONAL AUTHOR AFFILIATIONS
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49th Annual Meeting of the Orthopaedic Research Society
Paper #0122