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SIMULTANEOUS PREDICTON OF MUSCLE AND CONTACT FORCES IN THE KNEE DURING GAIT

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INTRODUCTION

Walking is important for human health, and independent ambulation predicts quality of life [1]. The study and treatment of neurological and joint disorders that inhibit walking would be more effective if muscle and joint forces could be determined reliably for individual patients. Knowledge of muscle forces is needed to characterize muscle coordination, which is a factor in neurological disorders such as cerebral palsy and stroke, while knowledge of joint contact forces is needed to characterize articular loading, which is a factor in bone and joint disorders such as osteoporosis and osteoarthritis. Reliable determination of these internal forces for individual patients would facilitate the design of customized surgical and rehabilitation treatments that maximize functional outcome.

Since clinical measurement of muscle and joint contact forces during walking is not feasible, musculoskeletal computer models have been the primary means for developing estimates. These models typically couple a dynamic skeletal model with individual muscle models but rarely include articular contact models due to their high computational cost. This omission makes estimation of muscle and joint contact forces a two-step process. First, muscle forces are estimated using a musculoskeletal model without articular contact, and second, joint contact forces are estimated by applying the muscle forces to a separate articular contact model. The problem with this approach is that the two models are inconsistent, with the first model requiring assumptions about which degrees of freedom are controlled primarily by muscles with little contribution from contact forces.

This study takes a fundamentally different approach by predicting muscle and contact forces *simultaneously* in the knee during gait. A single model combining muscle, articular contact, and dynamic skeletal models is used to develop the predictions. Two contacts (medial and lateral) are modeled for the tibiofemoral (TF) joint and one contact for the patellofemoral (PF) joint. The high computational cost of contact models is eliminated by using "fast" surrogate contact modeling techniques [2]. The muscle and contact force predictions are evaluated using *in vivo* tibial contact force measurements obtained from a patient implanted with a force-measuring knee replacement.

METHODS

Video motion, ground reaction, and tibial contact force data were collected from a single patient with an instrumented knee implant [3]. Institutional review board approval and informed consent were obtained. The patient's implant utilized a custom tibial prosthesis instrumented with four uniaxial force transducers, a microtransmitter, and an antenna. The patient performed isolated joint motion and gait trials, and one representative gait trial was selected for predicting muscle and contact forces simultaneously in the implanted knee.

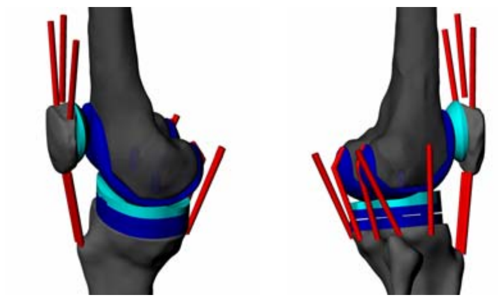

Fig. 1: Combined muscle, articular contact, and dynamic skeletal model used to predict muscle and contact forces simultaneously.

A composite muscle, articular contact, and dynamic skeletal model (Fig. 1) was constructed to develop the predictions. First, a

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patient-specific dynamic skeletal model was created from the video motion and ground reaction data [4]. Optimization methods were used to calibrate the joint and inertial parameters in the model based on data from the isolated joint motion and gait trials. Second, a geometric implant-bone leg model was created from CT data collected from the patient, implant CAD models provided by the manufacturer, and MRI data collected from a different subject of comparable stature. The bone and metallic implant components were segmented from the patient's CT data, and the implant CAD models and MR-derived bone models with muscle attachment points were registered to the segmented points. Third, the ankle, knee, and hip joint centers and functional knee axis in the dynamic skeletal model were registered to the joint centers and functional knee axis in the implant-bone model.

The combined model was used to predict muscle and contact forces simultaneously using inverse dynamic optimization methods. Eleven muscles crossing the knee and the PF ligament were included in the model. Muscle force was modeled as the product of activation level and the muscle's peak isometric force. Articular contact in the TF (medial and lateral) and PF joint was modeled using "fast" surrogate modeling methods [2]. Two cost functions were investigated. The first minimized the sum of squares of muscle activations [5], while the second minimized the sum of the three compressive contact forces [6]. Both cost functions were used with four different constraint sets composed of inverse dynamic loads for the three prescribed TF DOFs (Table 1). Each optimization function evaluation determined the static configuration of the remaining nine DOFs (i.e., three for the TF joint and six for the PF joint) given the current guess at the muscle activations, the calculated articular contact forces, and the inverse dynamic loads for the three free TF DOFs.

Constraint Set	Flexion-Extension Torque	Anterior-Posterior Force	Internal-External Torque
1	x		
2	x	x	
3	x		x
4	x	x	x

Table 1: Tibiofemoral inverse dynamic loads used as constraints.

RESULTS

The two cost functions predicted very similar muscle and contact force results, whereas the four constraint sets produced notably different results (Fig. 2). Medial contact force predictions were closest to the *in vivo* measurements when the flexion-extension torque was the only constraint, while lateral contact force predictions were closest when the flexion-extension and internal-external torques were used. Patellar contact force predictions were similar for all four constraint sets. When the *in vivo* medial and lateral contact force measurements were added as additional constraints, the model was able to match them closely, indicating that the muscle lines of action possessed sufficiently dimensionality to reproduce the *in vivo* conditions.

DISCUSSION

This is the first study to predict muscle and contact forces *simultaneously* in the knee during gait using a model that accounts for TF and PF contact. No assumptions were required about which inverse dynamic loads in the TF joint were controlled primarily by muscles, and no assumptions were required to determine the quadriceps moment arms. Availability of *in vivo* tibial contact force data provided a unique opportunity to evaluate the predictions. The best overall match to the medial and lateral contact force data occurred when the flexion-extension and internal-external torques from inverse dynamics were used as constraints. Our results suggest that important anatomic structures (e.g., small muscles, ligaments) or physiological features (e.g., muscle force-length and force-velocity properties) may be missing from the model, the wrong optimization cost functions may have been used, and/or a non-optimization (e.g. EMG-driven) method may be needed to predict *in vivo* contact force data accurately.

This study possesses several important limitations. First, ligaments were not included in the model, and it is possible that ligaments contribute significantly to some inverse dynamic loads. Second, only a single gait trial from a single subject was analyzed. It is rare to have access to gait and instrumented knee implant data collected simultaneously from the same patient. Nonetheless, data from additional trials and patients is needed to assess the extent to which these results are generalizable. Third, the patient used in this study had an implanted knee, so it is not known how well these results apply to subjects with natural knees.

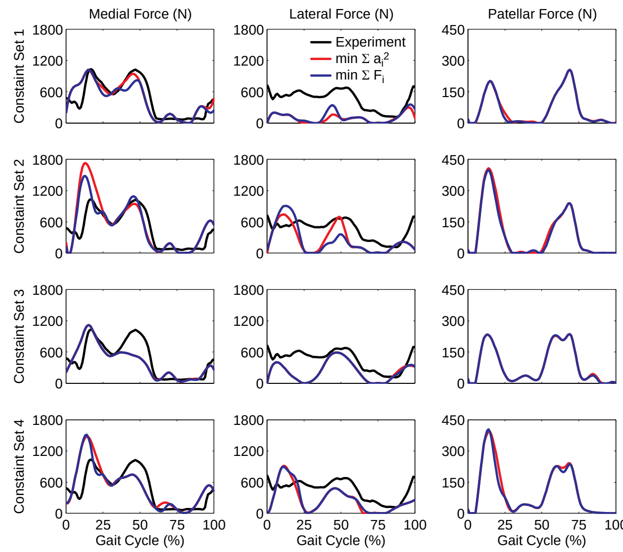

Fig. 2: Predicted contact forces for both cost functions and the four different constraint sets.

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REFERENCES

1. Muren, M.A., Hütler, M., and Hooper, J., 2008, "Functional capacity and health-related quality of life in individuals post stroke," Topics in Stroke Rehabilitation, Vol. 15, pp. 51-58.
2. Lin, Y.C., Haftka, R.T., Queipo, N.V., and Fregly, B.J., 2009, "Three-dimensional surrogate contact models for computationally-efficient multibody dynamic simulations," International Journal for Numerical Methods in Engineering (submitted).
3. Zhao, D., Banks, S.A., Mitchell, K.H., D'Lima, D.D., Colwell, C.W., and Fregly, B.J., 2007, "Correlation between the knee adduction torque and medial contact force for a variety of gait patterns," Journal of Orthopaedic Research, Vol. 25, pp. 789-797.
4. Reinbolt, J.A., Haftka, R.T., Chmielewski, T.L., and Fregly, B.J., 2008, "A computational framework to predict post-treatment outcome for gait-related disorders," Medical Engineering & Physics, Vol. 30, pp. 434-443.
5. Kaufman, K.R., An, K.W., Litchy, W.J., and Chao, E.Y., 1991, "Physiological prediction of muscle forces--I. Theoretical formulation," Neuroscience Letters, Vol. 40, pp. 781-792.
6. Schultz, A.B. and Andersson, G.B.J., 1981, "Analysis of loads on the lumbar spine," Spine, Vol. 6, pp. 76-82.

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