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MUSCLE AND CONTACT CONTRIBUTIONS TO INVERSE DYNAMIC KNEE LOADS DURING GAIT

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INTRODUCTION

Musculoskeletal computer models capable of predicting muscle and joint contact forces accurately during human movement could facilitate the design of improved joint replacements and new clinical treatments for articular cartilage defects or movement-related disorders [1]. A primary challenge to developing such predictions is the non-uniqueness of the calculated muscle forces, often referred to as the "muscle redundancy problem" [2]. Since more muscles act on the skeleton than the number of degrees of freedom in the skeleton, an infinite number of possible muscle force solutions exist.

The most common approach for predicting muscle forces involves inverse dynamic optimization [3]. Given movement data collected from a patient, the net reaction loads (i.e., three forces and three torques) acting at each joint are calculated using an inverse dynamic skeletal model. Muscle forces are then estimated by solving an optimization problem that minimizes a specified cost function (e.g., sum of muscle activations [2]) subject to constraints that muscles balance some (but not all) of the net loads. In physiological terms, these net loads are produced by a combination of muscle and contact (and possibly ligament) forces. Since contact forces are normally not modeled explicitly, assumptions need to be made about which inverse dynamic loads are balanced primarily by muscles (i.e., little contact contribution). These loads are the only ones that can be included as constraints in the optimization problem formulation.

This study uses in vivo data collected from a patient with a force-measuring knee replacement to quantify how muscle and contact forces contribute to inverse dynamic knee loads during gait. The results provide guidelines for which of the six net loads at the knee are balanced primarily by muscles, and thus which loads are "fair game" for use as constraints in muscle force optimizations if contact forces are not modeled explicitly.

METHODS

Video motion, ground reaction, and internal tibial contact force data were collected from a single patient with an instrumeted knee implant performing overground gait [4]. Institutional review board approval was obtained and the patient gave informed consent. The patient's implant utilized a custom tibial prosthesis instrumented with four uniaxial force transducers, a microtransmitter, and an antenna. Isolated joint motion trials were collected to determine joint functional axes, and one representative gait trial was selected for inverse dynamic and contact analysis.

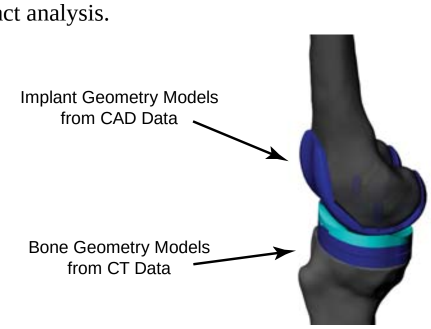

Fig. 1: Composite implant-bone model used to calculate contact contributions to inverse dynamic knee loads.

The video motion and ground reaction data were used to create a patient-specific inverse dynamic model and calculate six inverse dynamic knee loads [5]. In brief, the patient's skeletal structure was modeled as a parametric 14-segment, 27 degree-of-freedom full-body dynamic model. Joint parameters (i.e., positions and orientations of lower extremity joints in the body segments) and inertial parameters (i.e., masses, mass centers, and moments of inertia of the body segments) were determined from the isolated joint motion trials and gait trial using optimization methods. A foot-shank model utilizing the same parameter values was then used to calculate six inverse dynamic loads in the implanted knee.

The tibial contact force data were used to create a composite implant-bone model (Fig. 1) and calculate contact contributions to the six inverse dynamic knee loads. Geometric models of the patient's implant components were registered to geometric models of the patient's tibia and femur using CT data, and the composite implant-bone model was registered to the functional joint centers of the inverse dynamic model. Knee motion was specified using a combination of video motion measurements and separate fluoroscopic motion measurements performed on the same patient during treadmill gait. Contact between the femoral component and tibial insert was modeled using an elastic foundation contact model [6]. Small adjustments to superior-inferior translation, varus-valgus rotation, and medial-lateral translation were made via optimization such that the contact model reproduced the medial and lateral contact forces measured in vivo from the subject. Contact contributions to the six net inverse dynamic knee loads were then calculated, and muscle contributions were determined by subtracting contact contributions from the net loads.

RESULTS

Muscles were found to be the primary contributors to some net knee loads but not others. Maximum (average) differences between net and muscle loads across the gait cycle were 264 (150) N, 1636 (901) N, 52 (19) N, 9 (4) Nm, 2 (1) Nm, and 1 (0) Nm for anterior-posterior force, superior-inferior force, medial-lateral force, varus-valgus torque, internal-external torque, and flexion-extension torque, respectively. Thus, while contact forces had significant contributions to all three force components and varus-valgus torque, muscles (and possibly ligaments) were the primary contributors to internal-external torque and flexion-extension torque.

DISCUSSION

These findings support the use of the net flexion-extension torque and the net internal-external torque as constraints in optimization-based muscle force predictions that do not model contact forces explicitly. To reduce the amount of indeterminacy, one should consider using additional net loads as constraints only if a contact model is included explicitly in the musculoskeletal model used to predict muscle forces.

Two important caveats should be considered when evaluating these findings. First, the knee joint axis used in the inverse dynamic and implant-bone models was found via optimization methods. If the knee joint axis was estimated rather than optimized, contact forces would likely have had a significant contribution to the calculated inverse dynamic flexion-extension torque. Second, the rotation sequence used was different from the standard Grood and Suntay sequence [7]. While the flexion-extension axis was fixed in the femur, the varus-valgus axis (rather than the internal-external rotation axis) was fixed in the tibia so that the associated torque component would represent the knee adduction torque commonly used as an indicator of medial contact force. Given the small amount of varus-valgus and internal-external rotation in the composite knee model (only a few degrees), we do not expect that changing the order of the last two rotations would significantly alter the results.

This study also 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.

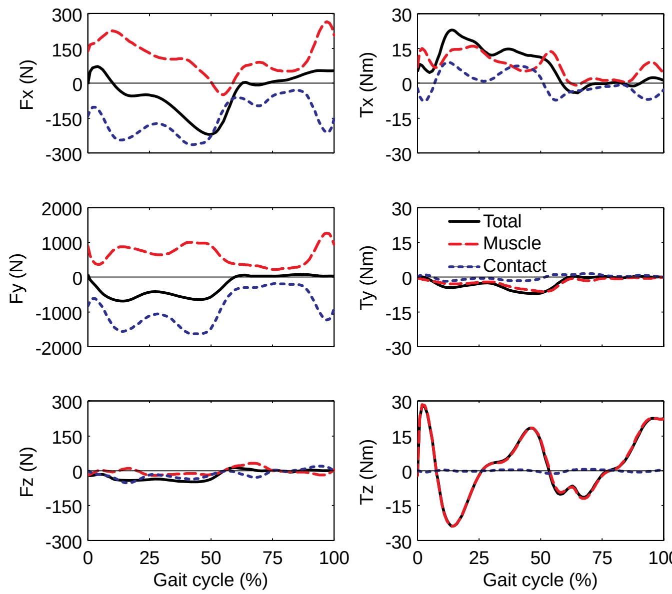

Fig. 2: Muscle (dashed red lines) and contact (dotted blue lines) contributions to inverse dynamic knee loads (solid black lines) over one gait cycle. F indicates force, T indicates torque, x direction is anterior-posterior, y direction is superior-inferior, and z direction is medial-lateral.

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