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EVALUATION OF PREDICTED KNEE JOINT MUSCLE FORCES DURING GAIT USING AN INSTRUMENTED KNEE IMPLANT

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INTRODUCTION

Many studies have used musculoskeletal models to predict *in vivo* muscle forces at the knee during gait [1, 2]. Unfortunately, quantitative assessment of the model calculations is often impracticable. Various indirect methods have been used to evaluate the accuracy of model predictions, including comparisons against measurements of muscle activity, joint kinematics, ground reaction forces, and joint moments. In a recent study, an instrumented hip implant was used to validate calculations of hip contact forces directly [3]. The same model was subsequently used to validate model calculations of tibiofemoral loading during gait [4]. Instrumented knee implants have also been used in *in vitro* and *in vivo* studies to quantify differences in biomechanical performance between various TKR designs [5, 6]. The main aim of the present study was to evaluate model predictions of knee muscle forces by direct comparison with measurements obtained from an instrumented knee implant. Calculations of muscle and joint-contact loading were performed for level walking at slow, normal, and fast speeds.

METHODS

A series of experimental and computational steps were followed to evaluate model predictions of *in vivo* tibiofemoral contact forces during gait. First, experimental gait and imaging data were recorded from a male subject (age 80; mass 68 kg; height 1.7 m) who wore an instrumented knee implant on his right leg. *In vivo* tibiofemoral forces were recorded simultaneously with video motion and ground reaction forces for three speeds of overground walking: normal (1.24 ± 0.03 m/s), fast (1.52 ± 0.04 m/s), and slow (0.80 ± 0.05 m/s). Data were recorded for three trials at each gait speed. Four uniaxial load cells imbedded in the implant provided a measure of the total tibiofemoral contact force [7]. The load cell measurements were converted into medial and lateral contact forces using regression equations reported by Zhao et al. [8]. Single-plane fluoroscopic data were also recorded as the subject walked on a treadmill at the same speeds used in the overground trials. Shape matching techniques were used to quantify the motion of the femoral component and tibial tray. Second, video motion and ground reaction forces measured in the gait experiments were input into a patient-specific musculoskeletal model of the lower limb to calculate leg-muscle forces using inverse dynamics and static optimization techniques. The body was modeled as an 8-segment, 21-degree-of-freedom articulated linkage, actuated by 58 Hill-type muscles. The performance criterion was to minimize the sum of the squares of the muscle activations in the model [9]. Third, composite implant/bone/muscle geometric models were created using computed tomography (CT) data recorded from the TKR patient, magnetic resonance imaging (MRI) data recorded from a healthy subject similar in stature to the TKR patient, and computer-aided design (CAD) models of the knee implant components. Accurate CAD models of the implant components were registered to the segmented post-surgery bone/implant CT data using the segmented implant geometry. Muscle geometries obtained from MR were also registered to the bone geometries. Fourth, the implant/bone models and modeled muscle paths were used to create a finite-element (FE) joint contact model of the knee. Measured values of tibiofemoral bone motion obtained from fluoroscopy and ground reaction forces, together with the calculated values of lower-limb muscle forces, were then input into the FE model to obtain total tibiofemoral contact force, as well as the forces transmitted by the medial and lateral sides of the knee.

RESULTS

There was no statistical difference between the measured and calculated total tibiofemoral contact forces for each of the gait trials

($p>0.05$); however, the measured and calculated values of the medial and lateral tibiofemoral contact forces were statistically different ($p<0.05$). Qualitatively, the medial contact force was over-estimated in the model, while the lateral contact force was under-estimated. For all 9 walking trials, the mean error in the calculated total tibiofemoral contact force was 162.0 ± 165.4 N, while the errors in the medial and lateral contact forces were 173.3 ± 113.3 N and 165.2 ± 134.2 N, respectively. All errors were calculated over one full cycle of walking. Exemplar results for the fast walking speed are shown in Fig. 1. The temporal behavior of the calculated muscle forces was consistent with stereotypical EMG data reported in the literature [1,7].

DISCUSSION

To our knowledge, this study represents the first attempt to evaluate model calculations of knee muscle forces using an instrumented total knee replacement. Lower-limb muscle forces were found by combining a detailed musculoskeletal model with inverse dynamics and static optimization theory. Measurements of joint kinematics and the ground reaction force, together with the calculated values of leg muscle forces were then input into a finite-element model of the knee to obtain tibiofemoral joint loading during gait. The model calculations were quantitatively compared to tibiofemoral joint contact forces measured in the same subject.. The results showed that the modeling framework predicted the total and medial contact forces well, although differences between the measured and calculated peak values were variable between the nine walking trials (see Fig. 1). The lateral contact forces were predicted less accurately. Furthermore, tibiofemoral contact forces were predicted well during the swing phase, but the errors fluctuated in stance. Despite these discrepancies, our findings suggest that *in vivo* knee-joint loading can be estimated with reasonable accuracy using muscle forces obtained from musculoskeletal modeling and optimization theory.

There are a number of important limitations of our work, each of which may have contributed to the discrepancies observed between model and experiment. First, anterior-posterior translations, internal-external rotations, and flexion-extension movements of the knee were all recorded for treadmill walking using single-plane fluoroscopy, and these kinematic measurements may not have been compatible with those obtained from the same patient for overground gait. Second, the muscle origin and insertion sites, and therefore the calculated values of muscle moment arms in the model, were determined using bone geometry from a different subject. Third, the values assumed for the musculotendon parameters in the model were taken from a generic model of the body [1] and were not calibrated to the patient used in this study. Finally, no ligament or soft tissue forces were included in either the musculoskeletal model or the finite-element model of the knee.

The model calculations were particularly sensitive to the moment arms assumed for the biceps femoris long head and patella tendon and to the insertion site assumed for the quadriceps muscle. Future work should be directed at estimating these quantities as accurately as possible, because the hamstrings, quadriceps, and patellar tendon provide the major resistance to the adduction moment at the knee created by the external ground reaction force. The model also included a bound on the tibiofemoral contact centre of pressure (CoP), which proved to be an important constraint, as this quantity had a significant effect on the magnitudes of the medial and lateral contact forces calculated in the model.

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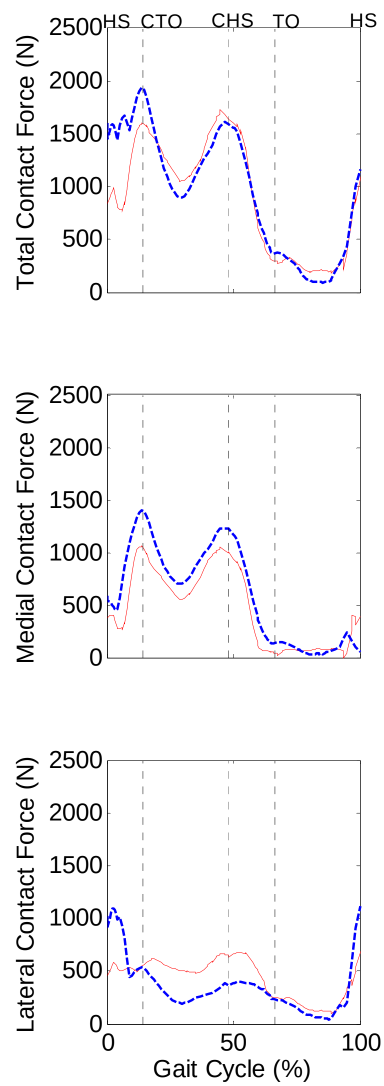

Fig. 1: Comparison between measured (solid lines, red) and calculated (dashed lines, blue) tibiofemoral contact forces for fast walking.