INTRODUCTION

The onset and progression of knee osteoarthritis is most often attributed to an injury or pathology that alters the load distribution at the tibiofemoral joint. A change in normal joint loading may be due to various factors, including damage to the soft tissues such as the knee ligaments and menisci, abnormal alignment of the tibia relative to the femur, musculoskeletal weakness, or some combination thereof. A number of studies have reported that surgical treatment of soft tissue injuries and/or correction of tibiofemoral alignment may decrease the progression of knee osteoarthritis. Relatively little is known, however, about how the knee ligaments, menisci, and muscles contribute to the distribution of medial and lateral compartmental loading at the tibiofemoral joint during activity. An overall goal of our on-going work is to describe and explain the interactions between the muscles, soft tissues, and bones at the knee in vivo. In this study, a musculoskeletal model of the lower limb is used to predict the distribution of tibiofemoral joint loading in normal gait.

No data currently exists to indicate medial and lateral compartmental loading at the tibiofemoral joint during activity. Only Morrison [1] and Harrington [2] have reported predictions of the resultant contact at the tibiofemoral joint and the location of its center of pressure during normal gait. The paucity of data on knee loading is due to the difficulties inherent in the measurement and calculation of force in vivo. In this study, tibiofemoral joint loading was obtained using a three-dimensional model of the lower limb, into which a detailed model of the knee was incorporated [3]. Muscle forces, ground-reaction forces, and joint motion obtained from a dynamic optimization solution for normal gait were used as input to the lower-limb model. We hypothesized that at any instant during stance more than one-half of the resultant force acting between the femur and tibia is transmitted by the medial condyles.

METHODS

In the lower-limb model, five rigid bodies were used to represent the right leg: thigh, patella, shank, hindfoot, and metatarsals. All joints were represented as described by Anderson and Pandy [4], except the knee, which was modeled as a six degree-of-freedom joint [3]. The geometry of the distal femur, proximal tibia, and patella was based on cadaver data reported for an average-size knee. The contacting surfaces of the femur and tibia were modeled as deformable, while those of the femur and patella were assumed to be rigid. The compressibility of the articular surfaces was adapted from measurements performed on cadavers. Thirteen elastic elements were used to describe the geometric and mechanical behavior of the knee ligaments and joint capsule. The properties of these elements in the model were adjusted to match the measured laxity of the tibiofemoral joint obtained from cadaver experiments. Thirteen muscles actuated the model of the lower limb.

Joint motion, ground-reaction forces, and muscle forces were obtained from a simulation of normal gait [4] in which muscle metabolic energy consumption was minimized over a full gait cycle. The gait simulation results were closely similar to measurements obtained from five subjects who walked at their self-selected speeds [4].

Inverse dynamics was used to determine joint-contact loading and the relative positions of the bones at the knee at each instant during the gait cycle. Specifically, the joint angles, ground forces, and muscle forces obtained from the walking simulation were applied to the lower-limb model, and a static equilibrium problem was then solved to find the anterior-posterior and medial-lateral translations, varus-valgus orientation, joint-contact forces, and ligament forces at the knee.

RESULTS

Peak total tibiofemoral force (1960 N) occurred at contralateral toe-off and was 2.8 times body weight (Figure 1). From heel-strike to toe-off, tibiofemoral contact force remained predominantly on the medial side. Peak contact force on the medial side was about 1650 N, or 2.4 times body weight. Peak force on the lateral condyle was much less (560 N, 0.8 body weight). Lateral contact force was zero just before single-leg stance (CTO). During swing, the distribution of load on the medial and lateral sides was approximately the same.

DISCUSSION

Our results agree, at least qualitatively, with previous estimates of tibiofemoral joint loading reported by others. We found, as did Morrison [1] and Harrington [2], that the center of pressure of the resultant contact force lies mainly on the medial side of the knee during the stance phase of normal gait. Peak tibiofemoral contact force predicted by our model is also similar to that reported in the literature [1,2,5]. The force in the lateral compartment was zero when adductor moment was maximum in the model. Peak adductor moment predicted by the simulation (3.8 Nm) was similar to that reported in the literature for the gait cycle [5]. Together, these results support the contention that disproportionate loading on the medial side of the knee in gait is due to the predominantly adducting transverse knee moment applied to the knee [5].

One limitation of the current model is that it does not account for the way in which the menisci distribute tibiofemoral compartmental loading. Adding a meniscus to the model may change our results quantitatively by placing the centers of pressure of the medial and lateral joint forces farther from the center of the knee. However, it will not alter the distribution of tibiofemoral compartmental loading predicted by the model for normal gait.

REFERENCES


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