DEVELOPMENT AND IMPLEMENTATION OF AN OPTIMIZATION-BASED MODELING OF WEIGHT-BEARING LOADS ON THE KNEE JOINT IN PAEDIATRICS

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INTRODUCTION

The amount of mechanical load being transferred through the knee by the articular surfaces plays a major role in the development of both joint cartilage and skeletal structure. The history of mechanical loading on the knee follows a roughly typical trajectory during child development, being constrained by the progression in motor abilities. From an epigenetic perspective, this typical progression optimizes developmental outcomes of joint structure and function. This optimal trajectory, however, can be disrupted by various factors, including malalignment of the mechanical axis of the lower limb, neuromuscular impairments, and cumulative high loads. As an example, there is some evidence that childhood obesity may alter the development of knee joint structures, resulting in a decreased accrual of articular cartilage[1]. This change in joint developmental outcomes could increase the risk for early onset of degenerative knee joint disease in this paediatric clinical population.

Measurement of the joint loading profiles on the knee would allow investigation of research hypotheses linking altered loading profiles with different developmental joint health outcomes. However, direct measurement of these internal loads requires invasive techniques. Bio-mechanical models can be used to provide indirect estimates.

Commonly, internal loading patterns are estimated from the external intersegmental moment which can be easily measured through motion capture and inverse dynamics calculation[2]. However, this resultant moment, along with the resultant rotation and flexion moments, includes the effect of all the internal structures of the joint, as well as muscles or ligaments spanning the joint. As such, a relationship between resultant external intersegmental moments and the magnitude of internal joint restraint forces, both shear and compressive, is difficult to establish for a general population.

Computational models have been developed to predict joint restraint and muscular forces in adults[3]. However, few attempts have been made at performing this type of analysis in children[4,5]. The development of a paediatric computational model would allow researchers and clinicians to predict the muscle activation and joint loading profiles in a variety of clinical populations. Boundaries of developmentally safe joint restraint forces could be established to help develop clinical guidelines, and non-invasive research into the effects of various neuromuscular and musculoskeletal conditions could be undertaken.

The aim of this study was to develop an optimization-based model of the lower limbs, test its sensitivity (tolerance to error), and explain the limits for its use in paediatrics.

METHODS

Skeletal Model

Skeletal landmarks were recorded during static standing trials and used to construct a linked-segment model using C-Motion (Visual3D Inc, Germantown, MD). Reference positions for all the landmarks were used to estimate joint centers and all of these locations were assumed to remain stationary within the local segmental coordinate systems.

Applying inverse dynamics to the skeletal system generated resultant intersegmental forces and moments, $\overrightarrow{F_{Res,j}}$ and $\overrightarrow{M_{Res,j}}$ respectively, which include the effects of all the structures crossing the *j*-th joint. These forces and moments, along with the motion of the rigid segments and landmarks, were exported to Matlab (Mathworks, Massachusetts, USA).

Kinematics

The Terry Musculoskeletal Database[6] provided the location of the origin, insertion, and via points for 42 independent muscle components. These locations were anisotropically scaled to the subject[7] and used to define a normalized muscular force vector, $\vec{v_{i,j}}$, from origin to insertion, for each muscle, *i*, crossing joint *j*[8]. An effective origin or insertion was used when a relevant via point was present. For the quadriceps, a cylindrical wrapping technique was implemented to replicate the effect of the patella[9].

The muscle moment arm, $\vec{r_{i,j}}$, was defined as the vector from the muscular insertion, or effective insertion, and the *j*-th joint center[8]. Thus, when combined with the unknown muscular force magnitude, f_i, the moment generated by each muscle was given by:

$$\overrightarrow{M_{i,j}} = \overrightarrow{r_{i,j}} \times \left(f_i \cdot \overrightarrow{v_{i,j}}\right) \qquad (1)$$

System of Equations

The resultant intersegmental moments at each joint, $\overrightarrow{M_{Res,j}}$, must be balanced by internal forces, including muscles, ligaments, and joint architecture. In this first approximation, the contribution of ligaments and other intra-articular joint structures was assumed to be negligible along axes of motion primarily controlled by muscle-generated moments: flexion/extension axes at the ankle and knee, and all three axes at the hip. This creates a system of 5 moment balance equations, as a function of our 42 unknown f_i values[3]:

$$\overrightarrow{M_{Res,\mu}} + \sum_{i=1}^{42} \overrightarrow{M_{i,\mu}} = 0 \quad , \qquad (2)$$

where μ represents the 5 previously specified axes used in the moment balance.

Optimization

An optimal solution to this indeterminate system[3] was found by minimizing total muscle stresses as the cost function[10]:

$$U = \sum_{i=1}^{42} \left(\frac{f_i}{A_i}\right)^2 \quad , \qquad (3)$$

where A_i is the scaled physiological cross sectional area (PCSA) of the *i*-th muscle component. The upper and lower limits of muscular force, f_i, were initially set to be strictly positive, with a maximum stress, σ_{max}, of 1 MPa[11]. To prevent non-physiological increases or decreases in muscle activation, upper and lower bounds were redefined at subsequent time frames based on the muscle activation levels at the previous iteration[3,12].

Joint Restraint Forces

To calculate the joint restraint forces (JRFs), the sum of the muscular forces acting around each joint were subtracted from the resultant intersegmental forces:

$$\overrightarrow{F_{Res,j}} + \sum_{i=1}^{42} f_i \cdot \overrightarrow{v_{i,j}} + \overrightarrow{F_{C,j}} = 0 \quad j \in \{1,2,3\} \qquad (4)$$

Experimental Methods

To test the model, data was collected on a representative 9 year-old male (height 132 cm, mass 35 kg). The child performed a sequence of 3 dynamic knee squats at a self-selected pace. The participant was instructed to stand upright, with feet shoulder width apart and arms folded over his chest, and to lower his body as low as possible, keeping his back straight, before returning to the upright position. The participant was oriented such that the body sagittal plane was aligned with the laboratory X-Z plane, with positive laboratory X anterior and positive Z proximal.

Segmental motions of the lower limb were collected at 100 Hz using two Optotrak 3020 position sensors (NDI, Waterloo, ON). Ground reaction forces (GRFs) were recorded for each lower limb at 1000 Hz using two AMTI BP400600NC-1000 force plates (AMTI, Watertown, MA, USA). Segment positions and GRFs were exported into C-Motion and low-pass filtered (3[rd] order Butterworth double-pass filter, with a cutoff at 6 Hz for position data and 10 Hz for GRFs).

Sensitivity Analysis

A sensitivity analysis was performed to determine effects of standardized changes in GRFs on model estimates. The recorded data of a single trial was modeled and used as the baseline prediction. The GRF magnitude was then modified by increments of ±5% and ±10% in each of the three laboratory coordinate system component directions. These 12 additional sets of GRF data were modeled using the original motion capture data.

RESULTS

The optimization routine converged to an optimal solution for all 3 trials, yielding similar joint force loading profiles. Knee compression, anterior shear and medial shear forces increased uniformly, reaching peak values at the base of the squat (highlighted in blue in Figure 1).

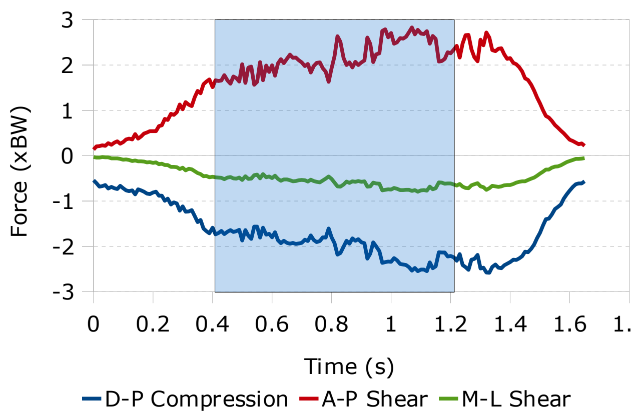

Figure 1: Representative Profile of Joint Restraint Forces For a Single Trial.

The following model predictions were calculated: average peak compressive force of 2.27 (SD ±0.32) times body weight (xBW), with a maximum of 2.58 xBW found in Trial 2; average peak anterior-posterior shear force of 2.32 xBW (SD ±0.55), with a maximum value of 2.82 xBW on Trial 2. Peak total resolved JCF values ranged from 2.7 to 3.5 xBW across the three trials.

Sensitivity Analysis

Deviating the GRF in the three laboratory coordinate system component directions produced significant changes in the joint restraint forces. Varying the magnitude in the anterior (X) direction produced small variations in the joint loading across the entire trial (Figure 3), and did not significantly alter the magnitude of the peak force magnitude (Figure 2). A modification in the (Y) direction resulted in more average error over the entire trial, but both the largest average error and peak magnitude error were found with a 10% increase in the laboratory vertical (Z) direction of the GRF. Error in both peak and overall magnitude was approximately proportional between the 5 and 10% tests.

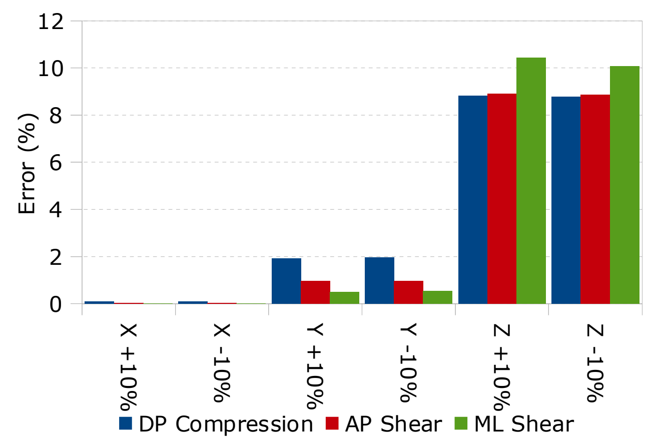

Figure 2: Percentage Error in Peak Loading With Modified GRF Components

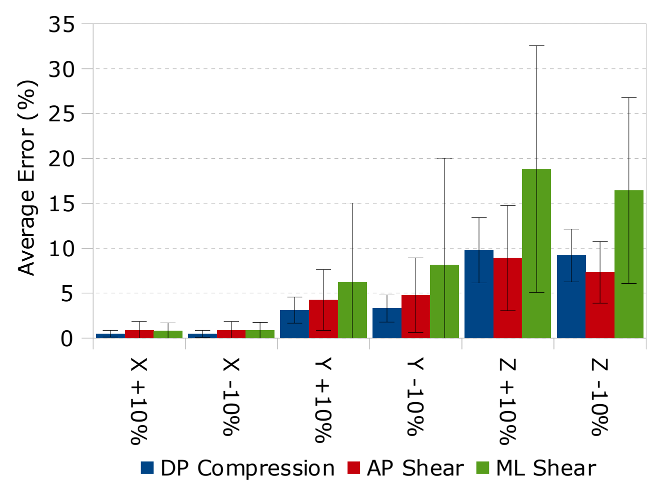

Figure 3: Average Percentage Error With Modified GRF Components

DISCUSSION

Model Performance

The model was found to converge uniformly across all three trials, indicating a high level of robustness in the scaled anthropometric model. Both the compressive forces and the anterior-posterior shear forces were consistent across all three trials, as evidenced by the low SD values.

Modifying the ground reaction force had a measurable effect on the joint loading. This indicates a sensitivity to errors and differences in GRF distribution, indicating that the model has the capacity to sense differences between loading patterns and adjust to accordingly. The model is, therefore, deemed to be sufficiently robust and sensitive to investigate knee loads in a diverse paediatric population.

Comparison to Reported Data

Few studies have looked at the deep squat using computational modeling, and even fewer have applied models to a paediatric population. Early planar modeling of slow squats produced peak compressive and anterior shear forces of 5.11 and 3.88 xBW respectively[13], while minimum-time squat-to-stand movements produced xBW loads 57% larger than in our paediatric subject[14]. These values are significantly higher than our predictions, but were restricted to the sagittal plane. A 3D model of a deep squat produced a total JRF across all three components between 1.8 and 3.0 xBW[15]. Since it has been shown that a scaled anthropometric adult model applied to children produces similar results, comparable values would be expected from our model[5]. However, across our three trials, the peak total JRF ranged from 2.7 to 3.5 xBW, higher than those values reported for the adult population.

Future Work

This computational model is a strong platform for designing more advanced versions, capable of studying the effects of neuromuscular and musculoskeletal conditions. Implementation of a patellofemoral contact force and ligaments would further refine the joint restraint force predictions, while incorporating paediatric muscle data would further tailor this model towards that population. Alternative optimization routines enforcing both co-contraction and muscle spasticity can be developed to study the effects of these conditions on the development of articular cartilage.

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