A 3D Computer Model For Non-Contact ACL Injury Simulations

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Introduction

Anterior cruciate ligament (ACL) injury is one of the most common and potentially traumatic sports related injuries. Approximately 80,000 injuries occur annually within the United States (Daniel and Fritschy, 1994). ACL injury also presents the potential for long-term morbidity, such as in the case of osteoarthritis (Maletius and Messner, 1999). Non-contact ACL injuries, accounting for approximately 70% of all ACL injuries, commonly occur during the landing or stance phase of movements incorporating sudden deceleration and or rapid speed or direction changes, such as sidestepping. Previous research has suggested a relationship between isolated variables linked to sidestepping (e.g., joint kinematics) and non-contact ACL injury (Cross et al., 1987; McLean et al., 1998). Research of this type however cannot provide information on forces generated within the ACL during these movements, and cannot be used to identify mechanisms of injury. The purpose of this study was to develop and validate a 3D computer model that could simulate sidestepping and estimate resultant ACL forces during these movements. This model was then used to determine the effects of variability in neuromuscular control during sidestepping on ACL injury risk.

Methods

Three-dimensional (3D) whole body and segment angular (J₁…J₉) and linear positions and velocities, and synchronous 3D ground reaction force (GRF) data (J₁₀…J₁₂) were measured in a single subject over ten sidestepping trials. Mean (±SD) segment positions and velocities at impact were quantified and submitted as initial conditions to a forward dynamics 3D musculoskeletal model. The 3D rigid-body model of the trunk and lower extremity consisted of four skeletal segments, two wobbling masses and 12 skeletal degrees of freedom (Figure 1). Wobbling masses were implemented to provide realistic impact responses. Equations of motion for forward dynamic simulation were generated by SD/FAST (Symbolic Dynamics). Models for the muscle forces and ground contact were added using custom functions. Neural excitation signals driving each muscle were modeled as step functions described by five parameters and were found by an optimization that minimized the difference between simulated (\(v^{sim}\)) and measured (\(v^{meas}\)) movement and ground reaction force variables from impact to 200ms using a simulated annealing algorithm (Equation 1).

\[
\text{minimize } f(p) = \sum_{j=1}^{12} \sum_{i=1}^{200} \left[ \frac{v^{sim}_j(t_i) - v^{meas}_j(t_i)}{SD_j} \right]^2
\]

(1)

The criterion for a good fit of the model was the difference between each measured and simulated variable should be less than two standard deviations, that is, \(f(p)\) was less than 9600. ACL force was derived for the optimized model from the A-P constraint force between the tibia and femur, combined with forces of the quadriceps and hamstring muscle groups acting across the joint (Herzog and Read, 1993). Monte Carlo simulations were performed (N=50000) to determine the effect of the measured variability in pre-impact body segment positions and velocities across the ten sidestepping trials on peak ACL force.

Figure 1. 3D lower extremity musculoskeletal model comprising four skeletal segments, two wobbling masses and 12 skeletal degrees of freedom.
Results

Simulated kinematic and GRF variables for the optimized solution were on average within 1.8 between-trial standard deviations of mean measured data (Figure 2), with the resultant peak ACL force estimated to be 108N, occurring at impact (Figure 3). Monte Carlo simulations produced a peak ACL force of $275 \pm 254$N, with 21 simulations demonstrating a peak force of greater than 1500N (Figure 4). These typically occurred with large hip flexion and small knee flexion values at initial contact.

**Figure 2.** Comparisons between simulated and measured variables for the optimized model of the sidestep for the first 100ms of stance.

**Figure 3.** ACL force estimations for the optimized 3D musculoskeletal model of the sidestep.
Discussion

Based on preliminary results, development and validation of a 3D model capable of simulating the stance phase of sidestep cutting maneuvers is possible. Monte Carlo simulation results suggest that variability in neuromuscular control prior to contact has a significant impact on the resultant force generated in the ACL during sidestepping. However, while a wide range of ACL forces were produced, the current model failed to generate ACL forces large enough to induce injury when realistic variability was incorporated. Limiting the estimation of ACL force to an A-P component is likely to underestimate the true ACL force, considering the complex 3D knee biomechanics that the sidestep has been shown to incorporate (McLean et al., 1998; Neptune et al., 1999). The inclusion of knee joint constraint torques (varus-valgus and axial rotation) is proposed for future model developments to overcome this current limitation (Kanamori et al., 2000). It is hoped that such improvements will provide a more realistic estimation of ACL forces during sidestepping and hence enable potential mechanisms of non-contact ACL injury to be identified.

References