INTRODUCTION: The anterolateral ligament (ALL) is a structure in the knee joint recently brought to attention [1]. Located obliquely on the lateral side of the knee, the ALL plays a role in stabilizing anterior-posterior (AP) and internal-external (IE) movements [2]. It has been reported that injuries to the anterior cruciate ligament (ACL) often coincide with the ALL injury [3], leading to altered knee movements [4]. Patients after ACL reconstruction along with concurrent treatment of the ALL reportedly experienced enhanced knee stability during the rehabilitation period [5]. According to a cadaveric experiment of sequential ligament removal and load testing using a robotic manipulator, significant improvement in rotational stability by the ALL was observed under static conditions [6]. However, there is a lack of studies on the effect of ALL injury on knee stability during dynamic situations. Recent advances in deep learning-based gait controllers allow forward dynamics simulations to generate desired behaviors [7]. The objective of this study is to investigate whether the combined injury of the ALL with an ACL-deficient knee leads to a significant increase in changes in AP and IE movements during walking using musculoskeletal forward dynamics simulation.

METHODS: We developed a full-body musculoskeletal model with 25 degrees-of-freedom (DOF) and 92 muscles in the lower limbs according to the full-body ‘Gait2392’ model [8] using the ‘RaiSim’ dynamics solver [9]. The muscle forces were calculated according to the Hill muscle equations in a previous study [11] and adjusted to match the cadaveric knee experiment [12]. Their stiffnesses and strains were adjusted by ‘Covariance matrix adaptation evolution strategy’ to reduce root mean square (RMS) kinematic error in various situations (three flexion angles and six loadings). Models of the injured knee were validated using a cadaveric experiment [6]. We prepared models for an intact knee, an ACL deficient (ACL-D) knee, and an ACL and ALL deficient (ACLALLD) knee. Quasi-static loading simulations were performed to quantify joint laxities for two loading cases at six knee flexion angles. A muscle controller for forward dynamics movement simulation of the intact knee model was trained using a reinforcement learning method [13] to closely imitate the measured gait kinematics of a subject (age 20 years, weight 63 kg, and height 1.73 m). Based on the kinematic state of the body, the muscle controller determined excitations of lower-limb muscles and torques of upper-limb joints at each frame. The same muscle controller was used for forward dynamics gait simulation of a fullbody model with unilateral ACLD and a fullbody model with unilateral ACLALLD. The knee kinematics of the intact, ACLD and ACLALLD models were observed for 100 gait cycles. A paired t-test with Holm adjustment was performed for statistical tests at a significance level of p < 0.05.

RESULTS SECTION: The root mean square errors of the knee model calibration for the six loading cases were less than 3 mm and 3 deg. Kinematic responses of the intact knee model to anterior force (Figure 1B), external torque (Figure 1C) and varus torque (Figure 1D) closely matched the cadaveric knee experiment. The ACLALLD model had larger anterior translation than the intact model by 8.6 mm, 7.6 mm, 7.9 mm, 8.5 mm, 7.8 mm, 1.4 mm at the knee flexion angles of 15°, 30°, 45°, 60°, 75°, 90°, respectively, and anterior force of 88 N (Figure 2A). The ACLALLD model also had larger internal rotation than the intact model by 6.1°, 3.6°, 4.0°, 3.9°, 4.5°, 6.4° at the flexion angles of 15°, 30°, 45°, 60°, 75°, 90°, respectively, and internal torque of 5 Nm (Figure 2B). During the stance phase, the peaks of the anterior tibial translation were 2.3 ± 0.5 mm for the intact knee, 4.7 ± 0.7 mm for the ACLD knee, and 4.6 ± 0.6 mm for the ACLALLD knee (Figure 3A). The anterior peak of the external tibial rotation in stance phase were 2.8 ± 0.5°, 1.3 ± 0.4°, -0.1 ± 0.6° for the intact, ACLD, and ACLALLD knees, respectively (Figure 3B).

DISCUSSION: Our knee models and forward dynamics simulations replicated clinical observations that patients with an ACL injury often exhibit increased anterior translation and reduced internal rotation during the stance phase [14]. The kinematic responses of our intact knee model to quasi-static passive loading were largely consistent with the joint kinematics in the cadaveric experiment, and were affected by ACL and ALL sectioning [6]. Prior research on the sectioning of the ALL in ACL-deficient knees suggested negligible impact on tibiofemoral translation [16], aligning with our findings that the ACLD and ACLALLD models showed no notable differences in AP translation during the stance phase. However, studies focused on the ACL reconstruction in ACL-reconstructed knee have reported corrected abnormal internal rotation in pivot shift test [17]. Our simulations revealed a significantly higher increase in internal tibial rotation for the ACLALLD model compared to the ACLD model, particularly during the mid-stance and terminal-stance phases. This suggests that, in dynamic contexts, the ALL plays pivotal role in regulating the IE rotation of an ACL deficient knee, providing tension toward external rotation during the stance phase [14].


IMAGES AND TABLES:

![Figure 1. Ligament structure (A) and kinematic response to anterior force (B), external torque (C) and varus torque (D) in 10 deg of the right knee flexion angle](image1.png)

![Figure 2. Validation of the injured knee’s response to anterior force (A) and internal torque (B)](image2.png)

![Figure 3. Anterior translation (A) and external rotation (B) during the stance phase of walking in the intact, ACL injured, ACL+ALL injured knee](image3.png)