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Short communication

# Dynamic contact stress patterns on the tibial plateaus during simulated gait: A novel application of normalized cross correlation

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ABSTRACT

The spatial distribution and pattern of local contact stresses within the knee joint during activities of daily living have not been fully investigated. The objective of this study was to determine if common contact stress patterns exist on the tibial plateaus of human knees during simulated gait. To test this hypothesis, we developed a novel normalized cross-correlation (NCC) algorithm and applied it to the contact stresses on the tibial plateaus of 12 human cadaveric knees subjected to multi-directional loads mimicking gait. The contact stress profiles at different locations on the tibial plateaus were compared, where regions with similar contact stress patterns were identified across specimens. Three consistent regional patterns were found, among them two most prominent contact stress patterns were shared by 9–12 of all the knees and the third pattern was shared by 6–8 knees. The first pattern was located at the posterior aspect of the medial tibial plateau and had a single peak stress that occurred during the early stance phase. The second pattern was located at the central-posterior aspects of the lateral plateau and consisted of two peak stresses coincident with the timing of peak axial force at early and late stance. The third pattern was found on the anterior aspect of cartilage-to-cartilage contact region on the medial plateau consisted of double peak stresses. The differences in the location and profile of the contact stress patterns suggest that the medial and lateral menisci function to carry load at different points in the gait cycle: with the posterior aspect of the medial meniscus consistently distributing load only during the early phase of stance, and the posterior aspect of the lateral meniscus consistently distributing load during both the early and late phases of stance. This novel approach can help identify abnormalities in knee contact mechanics and provide a better understanding of the mechanical pathways leading to post-traumatic osteoarthritis.

## 1. Introduction

Gait analysis based on stereo-photogrammetry (Cappozzo et al., 2005), dual-fluoroscopy (Li et al., 2008) and stereo-radiography (Tashman and Anderst, 2003) has been widely used to characterize in vivo joint kinematics. By combining knee joint kinematics with bony geometry, a connection between the tibiofemoral joint contact kinematics and the health of the cartilage at the site of contact has emerged (Anderst and Tashman, 2009; Beveridge et al., 2013). This has been further enhanced with the inclusion of articular cartilage into the models (DeFrate et al., 2004; Li et al., 2006). To date, an underlying but as yet unproven premise is that as the contact location changes after soft tissue injury (i.e., ACL rupture, meniscus tear), the spatial distribution and local characteristics of contact stresses on the articular cartilage also change, to which the tissue cannot readily adapt. To further explore this concept, a more detailed understanding of the regional contact stress patterns that articular cartilage is exposed to in uninjured knees is required. Unfortunately, patient-based studies do not allow for such measures, and most previous cadaveric studies do not mimic the complex multi-directional loading that knees experience in vivo. Added to this complexity is the fact that different regions of the articular cartilage are loaded and unloaded during different phases of activities of daily living (e.g., walking).

The objective of this study was to determine if common contact stress patterns exist on the tibial plateaus of human knees during gait. To satisfy this objective, we developed a novel normalized cross-correlation (NCC) algorithm and applied it to the contact stresses on the tibial plateaus of 12 human cadaveric knees subjected to multi-directional loads mimicking human gait.

## 2. Materials and methods

### 2.1. Experimental protocol

Twelve normal human cadaveric knees free of visible evidence of chondral defects, meniscus or ligament damage were used in this study (Table 1). The knees were carefully stripped of all soft tissue except for the capsule, collateral ligaments,

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cruciate ligaments and menisci. After stripping the knees, they were pinned through the epicondylar axis (Fig. 1c); this ensured that the flexion/extension (FE) axis of the knee was aligned with the FE axis of a load-controlled Stanmore Knee Simulator (University College London, Middlesex, UK) (Fig. 1a). The tibia–fibula complex and the femur were then potted into fixtures with polymethylmethacrylate (PMMA) bone cement. An electronic sensor (Model-4010N, Tekscan Inc., Boston, MA, USA) was placed on the tibial plateaus under the menisci and used to measure the contact stresses (Fig. 1b). The pressure sensor is a 22 × 34 matrix of sensing elements (sensel), and each sensel has a dimension of 2 mm × 2 mm. It was augmented with plastic tabs and sealed between two layers of Tegaderm adhesive dressing (3M, Minneapolis, MN) and calibrated with loads approximating 20% and 80% of the maximum axial load during gait (Brimacombe et al., 2009). The sensor was fixed to the surface of tibial plateau by suturing the augment tabs (Fig. 1b). Approximately 1 cm incisions were made in the meniscotibial ligaments anteriorly and posteriorly of both menisci, which allowed the sensor to be passed underneath the menisci with minimal disruption of the meniscocapsular attachments. The sensor tabs were then sutured in place using 3-0 Ethibond sutures via the tibial insertion of the anterior cruciate ligament (ACL) and the posteroinferior capsule (Bedi et al., 2010). The sensor position was adjusted to capture loads across the entire plateau under a 1000 N axial load (Fig. 1d). A custom scribe, which was attached through a drill hole in the cement mantle of the tibial potting block, was used to register positions of the sensor on the tibial plateaus ensuring the sensor was positioned consistently for each knee.

The simulator applied synchronized multidirectional loads (Fig. 2), including axial force, anterior–posterior (AP) force and internal/external (IE) torque to the tibia, while controlling the femur flexion/extension to mimic gait, according to ISO 14243-1. The other degrees of freedom of the tibia (medial-lateral translation, varus/valgus rotation) were uncontrolled (Bedi et al., 2010; Cottrell et al., 2008). To ensure reasonable, physiological joint kinematics, reflective markers were rigidly attached to the femoral and tibial fixtures for a sub-set of five knees (Fig. 1a), and their positions relative to the bone were registered with a 3D digitizer (accuracy: 0.23 mm) (MicroScribe; Immersion, San Jose, California). Anatomic landmarks (femoral epicondyles, medial/lateral edges of tibial plateau, tibial spine, etc.) were identified using the 3D digitizer to define the reference frames that describe the motion of the femur relative to the tibia (Wang and Zheng, 2010). The motion data were recorded at 50 Hz by a motion capture system (MotionAnalysis Inc., CA) and the normal contact stress was collected by a Tekscan sensor at 100 Hz. Twenty gait cycles were collected from each knee to ensure the sensor and the knee simulator reached steady state (Cottrell et al., 2008).

**Table 1**
Demographics of the knee joint donors.

| *Knee #* | *Side* | *Sex* | *Age* | *Weight (kg)* | *Height (m)* |
|---|---|---|---|---|---|
| 1 | R | F | 39 | 63.5 | 1.68 |
| 2 | L | M | 53 | 90.7 | 1.73 |
| 3 | R | F | 56 | 104.3 | 1.70 |
| 4 | R | M | 58 | 90.3 | 1.78 |
| 5 | R | F | 62 | 40.8 | 1.63 |
| 6 | L | F | 41 | 45.4 | 1.60 |
| 7 | R | M | 74 | 57.2 | 1.73 |
| 8 | L | M | 58 | 90.3 | 1.78 |
| 9 | L | F | 62 | 40.8 | 1.63 |
| 10 | R | M | 64 | 49.9 | 1.80 |
| 11 | L | F | 56 | 75.7 | 1.57 |
| 12 | L | M | 55 | 77.1 | 1.80 |
| Average ± SD | | | 56.5 ± 9.5 | 68.8 ± 22.2 | 1.70 ± 0.08 |

### 2.2. *Contact stress pattern within each knee*

To normalize the location of contact stress to individual tibia plateau geometry, the stress maps were aligned to the center of each meniscus and uniformly scaled based on the meniscus size. The meniscus was approximated by fitting a circle to the manually selected points on the periphery of each meniscus. This method was validated (medial $r^2=0.68$, lateral $r^2=0.72$, $p<0.01$, 5 knees) by comparing the radius of the best fitting circle to that measured from 3D meniscus models segmented from MR images (3D CUBE scan, voxel size: $0.3\times0.3\times0.6\ \text{mm}^3$) (Gold et al., 2007). The raw contact stress data at each sensel was then low-pass filtered (4th order zero-lag low-pass Butterworth) with a cut-off frequency of 6 Hz to remove high-frequency noise. The degree of similarity between the stress patterns of any two sensels was determined using cross-correlation, which has been widely used for pattern recognition in computer vision (Nakhmani and Tannenbaum, 2013). To remove the effect of contact stress magnitude on the pattern recognition algorithm, a normalized cross-correlation (NCC) algorithm was used (Eq. (1)). A

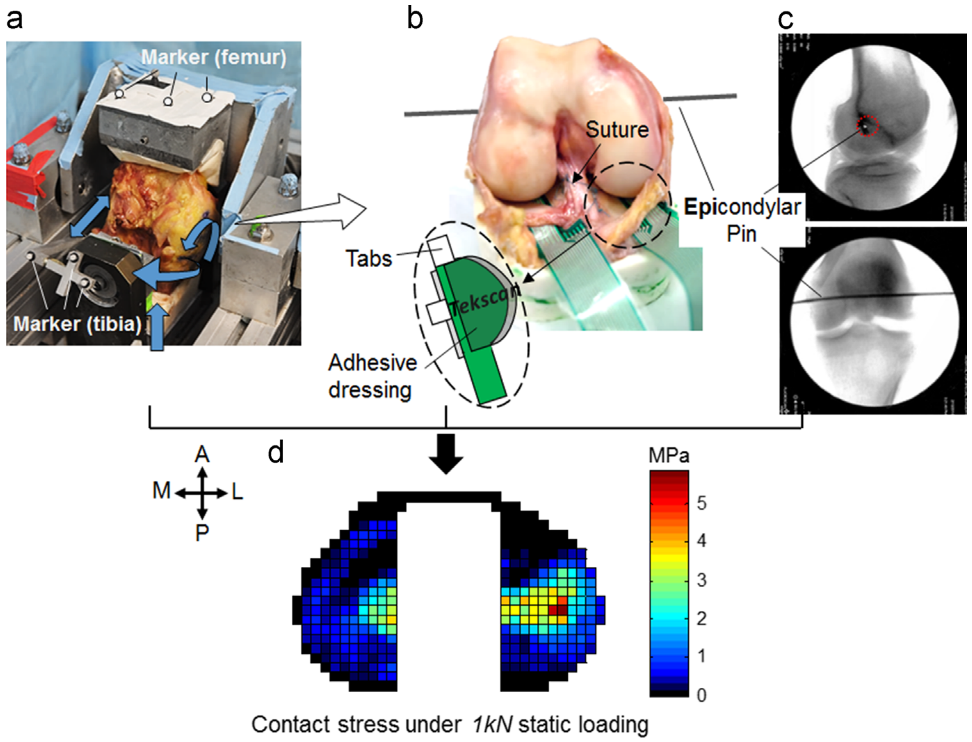

**Fig. 1.** Experimental setup for measuring the dynamic tibiofemoral joint contact stresses using a cadaveric model. (a) The femur and tibia were potted into fixtures with polymethylmethacrylate bone cement and mounted on a Stanmore knee simulator, the simulator applied dynamic axial forces, anterior–posterior (AP) forces, internal/external (IE) torques, and flexion/extension angle to mimic normal gait. (b) A Tekscan™ pressure sensor was augmented with plastic tabs and sealed between two layers of adhesive dressing. It was then placed on the tibial plateau by suturing the tabs to anterior cruciate ligament and posterior capsule. (c) Position of the femoral epicondyles in X-ray. (d) The contact stresses on the tibial plateau under 1000 N axial load.