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Subject-specific evaluation of patellofemoral joint biomechanics during functional activity

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ABSTRACT

Patellofemoral joint pain is a common problem experienced by active adults. However, relatively little is known about patellofemoral joint load and its distribution across the medial and lateral facets of the patella. In this study, biomechanical experiments and computational modeling were used to study patellofemoral contact mechanics in four healthy adults during stair ambulation. Subject-specific anatomical and gait data were recorded using magnetic resonance imaging, dynamic X-ray fluoroscopy, video motion capture, and multiple force platforms. From these data, *in vivo* tibiofemoral joint kinematics and knee muscle forces were computed and then applied to a deformable finite-element model of the patellofemoral joint. The contact force acting on the lateral facet of the patella was 4–6 times higher than that acting on the medial facet. The peak average patellofemoral contact stresses were 8.2 ± 1.0 MPa and 5.9 ± 1.3 MPa for the lateral and medial patellar facets, respectively. Peak normal compressive stress and peak octahedral shear stress occurred near toe-off of the contralateral leg and were higher on the lateral facet than the medial facet; furthermore, the peak compressive stress (11.5 ± 3.0 MPa) was higher than the peak octahedral shear stress (5.2 ± 0.9 MPa). The dominant stress pattern on the lateral patellar facet corresponded well to the location of maximum cartilage thickness. Higher loading of the lateral facet is also consistent with the clinical observation that the lateral compartment of the patellofemoral joint is more prone to osteoarthritis than the medial compartment. Predicted cartilage contact stress maps near contralateral toe-off showed three distinctly different patterns: peak stresses located on the lateral patellar facet; peak stresses located centrally between the medial and lateral patellar facets; and peak stresses located superiorly on both the medial and lateral patellar facets.

1. Introduction

Patellofemoral joint (PFJ) pain is a common problem experienced by active adolescents and adults, with incidence rates varying from 9% to 15% [1,2]. The condition is characterized by a gradual onset of peri-patellar pain and these symptoms are often aggravated by daily activities such as walking and stair ambulation [3]. In severe cases, PFJ pain can limit physical activity or result in its cessation altogether. Although the precise etiology of PFJ pain is currently unknown, high joint-contact stress, patellofemoral malalignment, quadriceps muscle weakness, and delayed onset of vastus medialis muscle activity are all thought to be contributing factors [4,5]. Importantly, it has been suggested that PFJ pain may be a precursor to the development of PFJ osteoarthritis (OA) [6], which is a common form of OA in the knee joint [7].

Mechanical loading of the patellar cartilage is believed to play a major role in the initiation and progression of PFJ pain symptoms. *In vitro* experiments [8,9] as well as rigid-body and deformable-body [10] computational models have been used to investigate PFJ biomechanics. *In vitro* experiments have provided useful insights, but these studies are limited by the fact that they do not replicate the levels of muscle and joint loading observed during daily activity. Similarly, rigid-body models are limited to the study of PFJ kinematics and contact forces. By comparison, deformable-body contact models are capable of calculating cartilage stress distribution, which is likely to be important in identifying factors contributing to structural joint deterioration [11].

Studies aimed at computing PFJ contact stress *in vivo* are scarce. Bretcher et al. [12] used a model that combined *in vivo* kinematic and kinetic data with contact areas calculated from magnetic

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resonance (MR) images and quadriceps moment arms obtained from the literature to estimate PFJ contact forces and stresses during stair ascent. They reported peak PFJ compressive forces and stresses on the order of 3.5 times body weight (BW) and 7 MPa, respectively. These results were derived using a simple model based on regression equations to estimate quadriceps muscle forces but contact stress maps were not computed. Chinkulprasert et al. [13] also used a combination of *in vivo* joint kinematic and kinetic data, muscle forces obtained from a regression model, and contact areas obtained from the literature to estimate PFJ forces and stresses during forward and lateral step-up and step-down tasks. Peak compressive forces and stresses for a forward step-down task were found to be 51.1 ± 2.7 N/kg and 13.8 ± 0.4 MPa, respectively. Besier et al. [14] developed a subject-specific deformable-body model of the PFJ to estimate patellar cartilage stresses *in vivo*. An open magnetic resonance (MR) imaging machine was used to measure joint kinematics, whilst muscle forces were estimated using an EMG-driven model. Farrokhi at al. [15] followed a similar approach to calculate joint contact stresses in both healthy subjects and patients with PFJ pain. A major limitation of each of these studies is that the kinematic and kinetic measurements were acquired at different times.

The purpose of the present study was to integrate existing capabilities in computational modeling with simultaneous measurements of *in vivo* joint kinematics and kinetics to estimate subject-specific PFJ kinematics, contact stresses, and stress distribution maps during weight-bearing activity. Stair ascent was investigated because this task is associated with higher PFJ loads than those present during either level walking or stair descent [12,16]. Our specific aims were firstly, to estimate the contact forces and stress distributions on the medial and lateral facets of the patellar cartilage during stair ascent; and secondly, to investigate the correlation between cartilage thickness maps and contact stress maps. We hypothesized that the locations of peak contact stress computed for the PFJ would correspond with the measured regions of maximum patellar cartilage thickness.

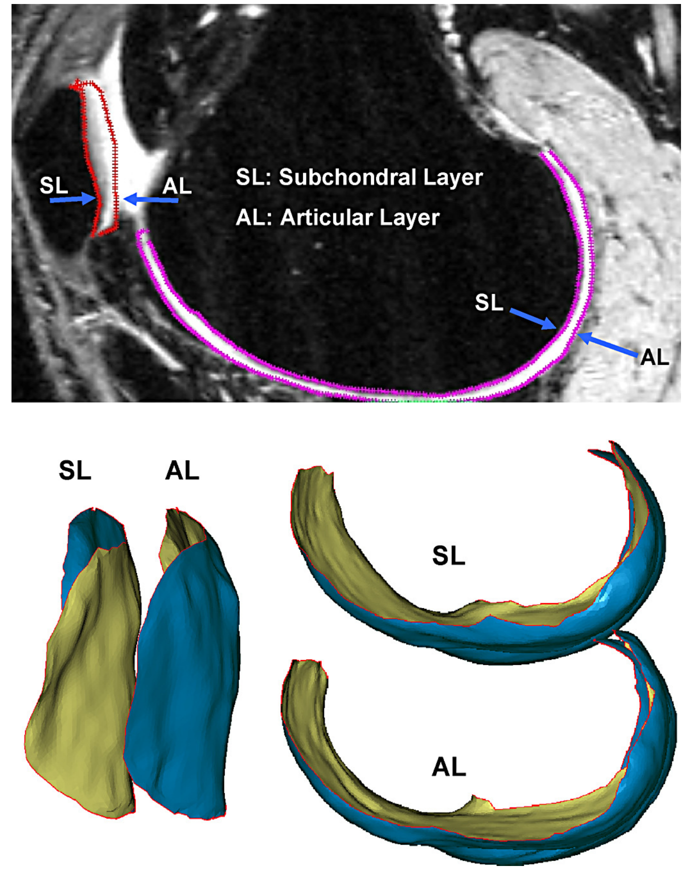

Fig. 1. Top panel: Magnetic resonance (MR) image illustrating how articular cartilage at the tibiofemoral and patellofemoral joints was segmented into two separate layers: a subchondral layer (SL) and an articular layer (AL). Bottom panel: Once the two layers were segmented from the MR images, surface meshes were wrapped to the segmented points on the femoral condyles and patellar facets.

2. Methods

2.1. Participants

Four healthy adult males (age, 30.5 ± 3 yrs; weight, 71.3 ± 7 kg; height, 178 ± 2 cm) with no history of lower-limb injury gave their informed consent to participate in the study after approval was obtained from the Human Research Ethics Committee at The University of Melbourne.

2.2. Magnetic resonance imaging

MR images of each subject's left lower-limb were acquired using two alternative MR sequences to obtain all the necessary anatomical information. The lower limb from the hip to the ankle was imaged using a T2 fat-suppressed sequence (TE = 12 ms, TR = 23 ms, NEX = 1, slice thickness = 1 mm × 1 mm × 1 mm). These images provided the bony geometries of the femur, tibia, fibula, and patella, as well as the origin and insertion locations of the quadriceps muscles. The patellar cartilage was separated into medial and lateral compartments using the patellar median ridge. A T2 fat-suppressed sequence (TE = 16 ms, TR = 29 ms, NEX = 1, resolution = 0.5 mm × 0.44 mm × 0.44 mm, FOV = 15 cm × 15 cm) using a knee coil was then obtained to visualize the articular cartilage. The MR images were segmented using a thresholding algorithm available in 3D Doctor (Able Software Corp, MA, USA) together with manual intervention. The geometry of the articular cartilage was segmented in two separate layers: the subchondral layer and the articular layer (Fig. 1). The surface geometries of the bones and articular cartilage were created from the segmented data points. The articular cartilage surfaces were registered to their corresponding subchondral bones using a rigid-body transformation that minimized the Euclidian distances between the surfaces of the bone and the subchondral layer of the articular cartilage. Patellar cartilage thickness maps were generated by calculating the normal distance in the anterior–posterior direction between the two articular cartilage layers using Geomagic (Research Triangle Park, NC). MR-based anatomical coordinate systems for the femur, tibia, and patella were defined as described in Fernandez et al. [17].

2.3. Gait experiments

Each subject performed a stair-ascent task on a custom-designed staircase comprised of four steps (step height = 31 cm; step width = 79 cm; step depth = 40 cm). Seventeen reflective markers were mounted on the subject's pelvis, left thigh, left shank, and left foot. A video motion capture system (VICON, Oxford Metrics Inc.) with nine cameras sampling at 120 Hz was used to record the marker-based kinematic data. Ground reaction forces were measured using two portable force platforms (AMTI Accugait, AMTI Corporation, Watertown, MA) mounted on the stairs. Muscle EMG data were recorded using a telemetry system (BIOTEL99, Neurodata, Vienna, Austria). Surface electrodes (Kendall Medi-trace 100, Tyco Healthcare Group) were placed over the rectus femoris, vastus medialis, and vastus lateralis muscles of the subject's left leg. EMG onset and offset times were determined by applying a