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# DETERMINING IN-VIVO HUMAN TIBIOFEMORAL CARTILAGE STIFFNESS USING DUAL FLUOROSCOPY AND MAGNETIC RESONANCE IMAGING

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## INTRODUCTION

Anterior cruciate ligament deficiency (ACLD) dramatically increases the risk of knee osteoarthritis (OA). Currently, there is no clinical diagnostic to predict joint degeneration in pre-radiographic OA. Cartilage stiffness, the resistance to deformation in response to an applied force, is a primary indicator of a joint's ability to resist compressive load and thus overall functional health. Methods for the in-vivo measurement of cartilage stiffness may lead to a promising functional diagnostic for populations at risk of developing OA. This research group uses high-speed dual fluoroscopy (DF), magnetic resonance (MR) imaging, three dimensional motion analysis, and computational modeling to quantify with sub-millimeter accuracy [1], [2] the in-vivo deformations of articulating tibiofemoral cartilages during standing load bearing. Preliminary results validate the accuracy and feasibility of measuring cartilage deformation and corresponding ground reaction forces. The goal of this research is to develop methods to combine these in-vivo deformations and joint force estimates to produce cartilage compressive stiffness estimates. This work supports the development of a new clinical diagnostic tool for pre-radiographic OA.

## METHODS

Preliminary data were collected for ACLD (n=4, >5 year history) and healthy control (n=5, CON) male subjects. The Conjoint Health Research Ethics Board (CHREB) approved the study, and informed consent was obtained for all subjects. Subjects were positioned in the field of view of the DF system, one of a limited number of systems globally that enables quantification of in-vivo six degree-of-freedom bone kinematics [3]. Subjects remained non-weight bearing for at least one hour prior to testing. Subjects were instructed to stand on the contralateral limb until instructed to shift their weight completely to the test limb. Standing weight bearing DF data for the test limb was collected at a rate of 6 Hz for 10 minutes (1 minute continuously and at 30 s intervals [2 s duration each] of the remaining 9 minutes). Force plates embedded within the instrumented treadmill (Bertec, USA) were used in conjunction with DF imaging to obtain the forces applied by the subject's legs to the ground (1200 Hz) during stance.

Subject-specific 3D bone and cartilage models were created in Amira (FES, Germany) from MR image data obtained for each subject immediately prior to the DF session (GE 3T Discovery 750, USA; Field of View 180 mm x 180 mm; 512 x 512 matrix; slice thickness 1 mm; 3D FIESTA sequence). DF Images were undistorted using a custom Matlab program (Matlab 2015b, MathWorks, USA) [4]. Bone kinematics were determined in Autoscoper (Brown University, USA) via a 2D to 3D matching approach [5] and applied to co-registered tibial and femoral cartilage models obtained from the MR images. Cartilage deformations were quantified as the change in median proximity of all model faces (surface normal distance to opposing surface) within 4 mm of the apposing surface (Matlab 2015b, Mathworks, USA).

## RESULTS

Cartilage Deformation

Median proximity data for one CON and two ACLD subjects (Figure 1) for the first 60 s of loading demonstrate substantially increased cartilage deformation for subjects ACLD 1 (1.33 mm) and ACLD2 (1.15 mm) compared to the CON subject (0.31 mm). The rate of deformation within the first 20 s was also much greater for the ACLD cartilages compared to the CON cartilages. Greater deformation indicates a reduced ability to resist compressive load. These results provide a proof of concept that tibiofemoral cartilage stiffness changes in pre-radiographic OA and that the in-vivo DF / MR measure is sensitive to detecting alterations in load deformation response.

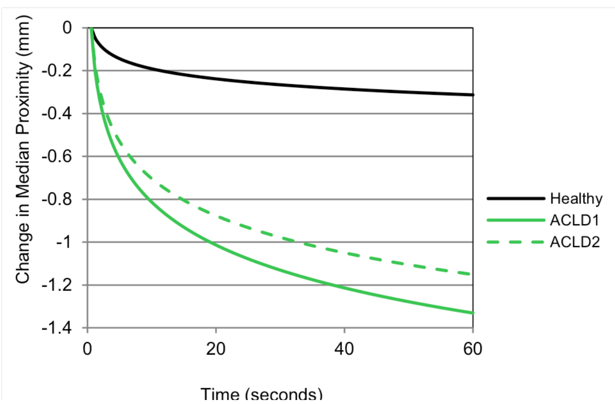


Figure 1: Change in median proximity of CON and ACLD subjects.

Ground Reaction Force

The vertical ground reaction force during the first 60 s of an exemplar standing weight-bearing trial (Figure 2), as applied by one subject to the force plate of the test limb, demonstrates fluctuations following initial loading. These are likely attributable to change in postural sway during the one-legged standing task. Initial full body weight loading appears to be completed within 2 s, with the initial positioning becoming stable within 10 s. The average force applied by the subject was 770 N ± 140 N.

It can be observed that the greatest rate of cartilage deformation occurs within the first 20 s for both healthy and ACLD subjects, despite full body weight loading being achieved almost instantaneously. This observed behaviour is consistent with the known viscoelastic properties of articular cartilage.

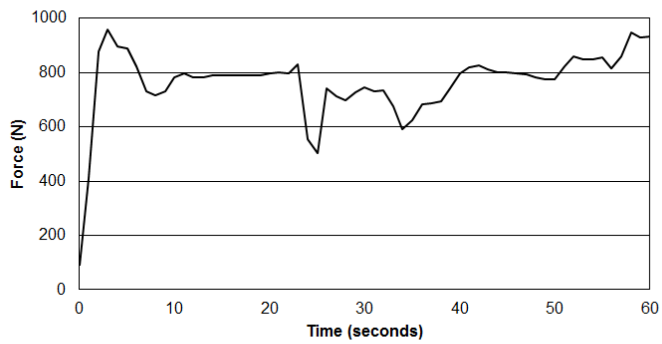


Figure 2: Measured vertical ground reaction force during one subject's static weight bearing test.

## LIMITATIONS

The viscoelastic nature of articular cartilage results in deformation behaviour that is strain-rate dependent. Cartilage tissue exhibits an instantaneous elastic deformation followed by a time-dependent long term creep behaviour when a load is applied [6]. In the current study, the subject controlled the loading rate himself, as he transferred body weight from the contralateral limb to the test limb. Although this pilot test allowed for preliminary understanding of the long-term creep behaviour of cartilage, the varying loading rate made analysis of instantaneous deformation response difficult.

## FUTURE WORK

To overcome some of the challenges associated with quantifying the instantaneous load deformation response, e.g. the high standard deviation observed from the loading curve, the test may be altered to load the subject's leg in a seated position with a custom knee loading apparatus (KLA) [7]. Subjects would be seated with leg fully extended to 0 degrees of knee flexion. In this position, the load and rate of application would be controlled and consistent between subjects. Through the use of established inverse dynamics techniques [8], [9], [10], [11] known applied loads would be translated to the knee joint. The resulting knee joint load would be plotted versus cartilage deformation and curve-fitting approaches would be used to determine material behaviour properties. Stiffness would then be calculated from the load versus deformation curve using the following formula:

$$k = \frac{F_j}{\Delta d} = \frac{\frac{\int_0^t F_j dt}{t}}{\Delta d} = \frac{1}{t\Delta d}\int_0^t F_j dt \qquad (1)$$

Obtaining an in-vivo model of cartilage stiffness mechanics will contribute toward the development of an early detection tool for pre-radiographic osteoarthritis, and enable longitudinal monitoring of joint health status.

## ACKNOWLEDGEMENTS

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