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A SAMPLE-SPECIFIC COMPUTATIONAL MODEL OF ARTICULAR CARTILAGE BASED ON MRI, HISTOLOGY, COMPUTER VISION AND MECHANICAL TESTING

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ABSTRACT

In a typical diarthrodial joint, like the human knee joint, the opposing bones are covered with a layer of dense connective tissue known as articular cartilage, which provides articulating surfaces. Articular cartilage has a composition and mechanical structure well-suited to the required functions: (i) to provide a compliant, low-friction surface between the relatively rigid bones in diarthrodial joints, (ii) to provide a long-wearing and resilient surface, and (iii) to distribute the contact pressure to the underlying bone structure. To meet these demands, articular cartilage contains a fluid phase of H_2O and electrolytes (approximately 68% to 85%), and a solid phase composed of chondrocytes, type I and II collagen fibers, proteoglycans and other glycoproteins (cf. [1]).

Within the cartilage, fibers of predominantly type II collagen exhibit a high level of structural organization and provide tensile reinforcement to the solid phase, a proteoglycan gel. The collagen fibers support only tension and accommodate essentially no resistance to compression. Within the cartilage, three basic zones of collagen fiber orientation exist. Starting from the surface, the superficial tangent zone (comprising approximately 10-20% of the total thickness) has fibers which are tangential to the articular surface. Next, the middle zone (40-60%) has fibers which are isotropically distributed and oriented. Finally, near the transition to subchondral bone, the deep zone (approximately 30%) has fibers which are oriented perpendicular to the aforementioned surface.

Clearly some simplifying assumptions are required to facilitate computational modeling and numerical simulation. By considering only two main components; a fluid and a solid embedded with type II collagen fibers, articular cartilage may be considered a biphasic fiber reinforced material [1]. Additionally, due in part to the interaction of solid and fluid, articular cartilage exhibits a time-dependent stress-strain response.

In light of the complexity of cartilage constitutive modeling, the motivation of our study is as follows: (i) to quantitatively relate the material properties and mechanical response of articular cartilage to the

underlying mechanical structure, (ii) to enable simulations of full joint, etc. with accurate stress estimates in the cartilage materials, (iii) to map the simulation results back to MRI data of the specific sample, and (iv) to enable further study of articular cartilage degeneration by numerical simulation.

Cartilage specimens were acquired from patients undergoing total knee replacement surgery[1]. After removal, the samples underwent a series of MRI sequences. MRI is widely accepted as a non-invasive technique for visualizing the morphology of healthy and degenerate articular cartilage [2]. A portion of the samples then underwent histological sectioning to capture the fiber density and orientation. The remaining portion was soaked in a bath of phosphate-buffered saline containing protease inhibitors (PBS+PI) in preparation for mechanical testing.

The imaging and mechanical testing were performed with a custom-built, robotics-based testing device (cf. [3] for details). After clamping the specimen, the cartilage surface was captured using a stereo camera setup. The testing device allowed interactive selection of the locations for the mechanical indentation on the reconstructed surface and automatic determination of the respective surface normals (the indentation directions). A stainless steel spherical tip (with 3 mm in diameter) was used to indent the cartilage specimen by 0.4 mm at selected locations. The deflection was held constant for 600 s, while the resulting load was recorded. To minimize the biomechanical effects of dehydration, the specimen was kept moist during the testing procedure by regularly spraying the surface with PBS+PI. After all mechanical testing has been performed, the cartilage layer was chemically stripped and the true bone surface was captured again using the stereo camera setup, yielding the full cartilage geometry.

To model the complex cartilage morphology and material response, the finite strain viscoelastic constitutive model, as documented in [4], is extended to include a biphasic formulation, creating a biphasic, viscoelastic fiber-reinforced constitutive response. This extended constitutive model is employed in a specimen-specific finite element (FE) simulation to recreate the force-displacement response of mechanical indentation tests where the contact algorithm from [5] is employed to capture specimen deformation due to indentation. Specialized analysis of patient-specific specimens (and a thorough literature search) is employed to determine the material parameters as well as the FE model geometry.

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