INTRODUCTION: The endplates of the intervertebral disc (IVD) distribute loads across the annulus fibrosus and nucleus pulposus (NP) structures of the disc, and have been shown to play an important role in the biomechanical behavior of the overall IVD. The endplates are known to deform under axial load [1], are thought to frequently fracture first in failure of the IVD, and are the primary path for fluid flow into and out of the IVD [2]. Human and animal bony and cartilaginous endplate properties have been measured using indentation testing [3] and permeation experiments [4]. However, it is difficult to implement the properties measured using these methods into constitutive models of the endplate and ultimately into finite element models of the IVD. In fact, the human endplate has been modeled as biphasic or poroelastic using properties measured from articular cartilage or animal endplate [e.g. 5-7]. With the advent of prosthetic devices for the IVD, including NP replacements that retain the endplate structures, it is crucial to accurately model the endplates to predict the performance of these devices. For instance, predicting endplate subsidence or a device’s access to fluid flow depends on accurately modeling the endplates using human endplate mechanical properties. Therefore, the objective of this study was to characterize the compressive biphasic mechanical properties of the human IVD cartilaginous endplate.

METHODS: Human lumbar spines were obtained under an IRB approved protocol. The average age of the donors was 56±23 y.o. (n=4). Bone-disc-bone units were obtained using a band saw, and the L2-L3 and L5-S1 endplate regions were removed from the bone-disc-bone unit was isolated to obtain cartilaginous endplate samples adjacent to the NP. Superior and inferior endplates were isolated from the bone-disc-bone units by removing as much NP and bone material as possible with a scalpel. A total of n=5 samples were taken from L2-L3 (n=4) and L5-S1 (n=1) endplates.

The endplates were cut into approximately 8 by 8 mm square samples. A freezing stage microtome was used to remove the rest of the NP and bone to obtain an endplate sample of uniform thickness. Sample thickness was measured using a laser sensor (±0.005mm), with sample thickness averaged over three points across the sample. The average thickness of the samples was 0.74±0.20 mm. A 4.37 mm diameter punch was used to create circular test samples. These test samples were placed within the confining chamber of a custom-built load- and displacement-controlled compression testing device.

A 0.05 N contact load was applied to the sample via a porous platen at 10 μm/second. The environmental chamber surrounding the sample was then filled with PBS and the sample was permitted to swell to equilibrium for up to three hours. Following this equilibrium period, a 10% compressive strain increment was applied at a rate of 250 μm/second and allowed to equilibrate for two hours. Data from this stress-relaxation curve was analyzed using linear elastic, isotropic biphasic theory [8]. Effective aggregate modulus, Hb, was defined as the ratio of the equilibrium stress to equilibrium strain. Apparent permeability (ka) was determined using linear biphasic theory to fit the stress-relaxation data [9].

RESULTS: The average apparent permeability was ka = 0.35±0.17 x 10^{-15} m^4/N-s. A representative stress vs. time dataset and corresponding biphasic curve fit are shown in Figure 1. Modeling studies of the IVD have used articular cartilage permeability, 0.5-10 x 10^{-15} m^4/N-s, to model human endplate. The permeability values measured here fall within that range. The permeability measured for baboon cartilaginous endplate, 14.3 x 10^{-14} m^4/N-s is orders of magnitude higher than that measured in this study [10]. The average aggregate modulus measured here for cartilaginous endplate was Hb = 0.82±0.48 MPa. This compares with values measured for baboon (0.44 MPa) and articular cartilage (0.2-1 MPa).

DISCUSSION: In this study, the biphasic mechanical properties of human cartilaginous endplate were measured for the first time. The permeability value falls within the range measured for articular cartilage, but was orders of magnitude lower than what has been reported for baboon endplate. The samples used in this study come from older donors whose tissue is likely degenerated, which likely decreases permeability. However, it is unlikely that degeneration causes an order-of-magnitude decrease in permeability. For example, degeneration decreases human NP permeability by 60%. A similar effect on endplate permeability would result in a value of 0.47 x 10^{-15} m^4/N-s for non-degenerated endplate (provided the value measured here was for degenerated tissue). Human NP permeability has been measured to be 0.9 and 1.44 x 10^{-15} m^4/N-s for non-degenerated and degenerated tissue, respectively [9], and to be 0.21 x 10^{-15} m^4/N-s in human annulus fibrosus, with no effect due to degeneration [11]. The lower permeability of endplate compared with NP is consistent with the function of the IVD, where fluid is pressurized within the NP as load is applied. The lower permeability of the endplate retains fluid within the IVD short-term, but ultimately allows fluid to flow out of the IVD to reach osmotic equilibrium, and to flow back into the IVD to repressurize. Though cartilaginous endplate permeability has been measured here to be similar to that of human annulus fibrosus, it is known that fluid flow is primarily through the endplates. The bony endplates adjacent to the endplates likely affect this preferential fluid flow.

The cartilaginous endplate is separated from the vertebral bone by the bony endplate. Studies have shown that the bony endplate is populated by narrow contact channels, which are holes through the bone endplate that make contact with capillary buds [12]. Nutrition and fluid flow to and from the cartilaginous endplate and hence, the IVD, mainly via these narrow contact channels. Therefore, fluid flow through the endplates is not only regulated by the permeability of the cartilaginous endplates, but also by the properties of the bony endplate. Studies have shown that narrow contact channel morphology and number change with IVD degeneration [12]. A larger sample size and a wider range of donor age than reported here is required to fully characterize the effect of age and degeneration on the properties of the cartilaginous endplate. Furthermore, bony endplate mechanical properties have been shown to vary spatially and with superior vs. inferior endplate [3], and these comparisons can be made for the cartilaginous endplates.

Modeling and disc explant studies have shown that the alterations to the permeability value of endplates adjacent to the IVD affect experimental and predicted mechanical behaviors of the overall IVD [5,13]. These studies demonstrate that correct input endplate mechanical properties are crucial for accurately modeling overall IVD mechanical behavior. The biphasic compressive human cartilaginous endplate mechanical properties measured for the first time in this study can be used to more accurately model endplate mechanical behavior, and hence, NP and overall IVD mechanical behavior.


AFFILIATED INSTITUTIONS: ** Drexel University, Philadelphia, PA; *** University of Pennsylvania, Philadelphia, PA.