INTRODUCTION
Clinical magnetization transfer (MT) sequences employ off-resonance saturation pulses followed by a conventional data acquisition 1-3. The MT pulse typically results in selective saturation of tightly bound water and collagen protons which exchange with the loosely bound water and then free water, leading to a loss of longitudinal magnetization and hence a signal reduction (Figure 1) 4. MT is ideal for probing interactions between protons bound to macromolecules and free water protons. Clinical MT sequences cannot detect MT effects in short $T_2$ tissues such as the menisci, ligaments, tendons and bone when there is little or no detectable signal present 5-6. In this study we evaluated ultrashort echo time (UTE) MT imaging of the meniscus. The angular dependence of MT ratio (MTR) as well as $T_2$ and $T_1$rho of meniscus were investigated.

MATERIALS AND METHODS
Four human knee menisci samples were harvested from cadavers. Each meniscus sample was subject to UTE-MT imaging as well as $T_2$ and $T_1$rho imaging using a 3 T GE whole-body scanner. The UTE-MT sequence was based on a regular 2D UTE sequence with a minimal nominal TE of 8 μs preceded by a MT pulse (a Fermi pulse with a duration of 8 ms). The 2D UTE-MT imaging protocol used the following parameters: TR = 300 ms, field of view (FOV) = 8 cm, matrix = 256×256, band width = 125 kHz, four echoes with TEs of 0.008, 4, 8 and 12 ms. The MT pulse was placed at ten off-resonance frequencies (1, 2, 3, 4, 5, 6, 7, 8, 9 and 10 kHz) with five different levels of MT power (300, 500, 700 and 1000). The UTE-MTs scans were repeated with each meniscus re-oriented at 10 different angles (0°, 20°, 40°, 60°, 80°, 100°, 120°, 140°, 160° and 180°) relative to the B0 field. $T_2$ measured with 2D clinical CPMG sequence (TEs of 10 to 80 ms) and $T_1$rho with a 2D spiral $T_1$rho sequence (spin locking time of 0 to 40 ms) were also performed 5. A home-build surface coil (~2.5 cm in diameter) was used for signal reception. Each meniscus sample was placed in a plastic container filled with perfluorooctyl bromide (PFOB) during MR imaging to maintain hydration and minimize susceptibility effects at air-tissue junctions. Images at different angular orientations were registered using a rigid-body model before quantitative analysis. The same ROIs were used for all subsequent MTR, $T_2$ and $T_1$rho calculations. MTR was plotted as a function of MT pulse frequency offset $\Delta f$, MT power $\theta$ and sample orientation.

RESULTS AND DISCUSSION
Figure 2 shows UTE-MT imaging of a meniscus at different frequency offsets and MT pulse powers. Clinical MT sequences show little signal from the meniscus. MTR values are difficult to assess with these sequences. The UTE-MT sequence provides high quality morphological images with high signal and resolution, as well as high quality MTR maps of the meniscus. Figure 3 shows UTE-MT of a meniscus sample at seven angular orientations (0° to 180°) and 10 different frequency offsets (1 to 10 kHz). The MT power was fixed at 1000. MTR was increased by 19% near the magic angle at 1 kHz frequency offset. MTR showed almost no angular dependence when the frequency offset was greater than 2 kHz. Figure 3B shows the angular dependence of MTR at four TEs of 8 μs, 4 ms, 8 ms and 12 ms with a fixed MT power of 1000 at 3 kHz. MTR decreased while the magic angle effect increased with longer TEs. While MTR showed nearly zero angular dependence for the first three TEs, a significant magic angle effect of 64% MTR increase was observed for the 4th echo.

Figure 4 shows the angular dependence of $T_1$rho of a meniscus sample. A maximal $T_1$rho value of 23 ms was observed, which is about 64% higher than the minimal $T_1$rho value of 14 ms. This result shows that $T_1$rho is more sensitive to the magic angle effect than $T_2$.

CONCLUSIONS
This study shows little magic angle effect for MTR but a strong magic angle effect for both $T_1$rho and $T_2$. These results suggest that UTE MT may be more robust in evaluating early OA.

REFERENCES