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Human talar and femoral cartilage have distinct mechanical properties near the articular surface

Corinne R. Henak [a,1], Keir A. Ross [b], Edward D. Bonnevie [c], Lisa A. Fortier [d], Itai Cohen [e], John G. Kennedy [b], Lawrence J. Bonassar [a,c,*]

[a] Meinig School of Biomedical Engineering, Cornell University, Ithaca, NY, United States
[b] Hospital for Special Surgery, New York, NY, United States
[c] Sibley School of Mechanical and Aerospace Engineering, Cornell University, Ithaca, NY, United States
[d] Department of Clinical Sciences, Cornell University, Ithaca, NY, United States
[e] Department of Physics, Cornell University, Ithaca, NY, United States

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ABSTRACT

Talar osteochondral lesions (OCL) frequently occur following injury. Surgical interventions such as femoral condyle allogeneic or autogenic osteochondral transplant (AOT) are often used to treat large talar OCL. Although AOT aims to achieve OCL repair by replacing damaged cartilage with mechanically matched cartilage, the spatially inhomogeneous material behavior of the talar dome and femoral donor sites have not been evaluated or compared. The objective of this study was to characterize the depth-dependent shear properties and friction behavior of human talar and donor-site femoral cartilage. To achieve this objective, depth-dependent shear modulus, depth-dependent energy dissipation and coefficient of friction were measured on osteochondral cores from the femur and talus. Differences between anatomical regions were pronounced near the articular surface, where the femur was softer, dissipated more energy and had a lower coefficient of friction than the talus. Conversely, shear modulus near the osteochondral interface was nearly indistinguishable between anatomical regions. Differences in energy dissipation, shear moduli and friction coefficients have implications for graft survival and host cartilage wear. When the biomechanical variation is combined with known biological variation, these data suggest the use of caution in transplanting cartilage from the femur to the talus. Where alternatives exist in the form of talar allograft, donor-recipient mechanical mismatch can be greatly reduced.

1. Introduction

Talar osteochondral lesions (OCL) have been reported with increasing frequency, due to higher resolution imaging and increasing sporting activity in an aging population. Most talar OCL result from a traumatic event, typically an ankle sprain. An estimated 50% of significant ankle sprains result in some form of cartilaginous injury (Ferkel and Chams, 2007; Saxena and Eakin, 2007; Takao et al., 2005). Of these OCL, 50% require some form of surgical intervention (Zengerink et al., 2010).

The long-term aim of OCL treatment is to restore the mechanical function of native talar cartilage, and to thereby prevent or delay progression of joint degeneration (Zengerink et al., 2010). To achieve this aim, two broad treatment options are available; either to repair the cartilage or replace it. While reparative strategies, including microfracture and microdrilling, have shown good short to medium term clinical outcomes, longer term outcomes in larger lesions are less promising (Savage-Elliott et al., 2014).

Larger talar OCL are typically treated by replacing defective cartilage and bone with an allogeneic or autologous osteochondral transplant (AOT), typically from the ipsilateral femoral condyle. Clinical outcomes following AOT have been excellent with functional scores of 90% or greater being reported in short and medium term outcome studies (Hangody et al., 2008; Valderrabano et al., 2009). Despite these excellent functional outcomes, concern remains regarding long term implications of mismatches in mechanical behavior between recipient and donor cartilage. Radiographic outcomes in several studies suggest that biologic integration between host and graft may not be optimal. For example, cyst formation in the subchondral bone has been shown

* Correspondence to: Cornell University, 149 Weill Hall, Ithaca, NY 14853, United States. Fax: +607 255 7330.
E-mail address: LB244@cornell.edu (L.J. Bonassar).
[1] Current address: Department of Mechanical Engineering, University of Wisconsin-Madison, Madison, WI, United States.

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in 67% of patients at a minimum of 43 months follow-up (Valderrabano et al., 2009), with biologic intervention via bone marrow aspirate insufficient to completely prevent their development (Kennedy and Murawski, 2011).

Further, previous cadaveric studies evaluating mechanical implications of graft placement have demonstrated that an elevated graft leads to increased contact pressure on the graft, while a graft of matched articular surface topology results in increased contact pressures in surrounding recipient cartilage (Fansa et al., 2011; Latt et al., 2011). This mismatch in cartilage contact mechanics may result in part from a fundamental mismatch in cartilage mechanical behavior between donor and graft cartilage. Recently, greater attention has being given to allogenic osteochondral grafts in the treatment of OCL (Görtz et al., 2010; Janis et al., 2010). This is in part due to the concern of donor site morbidity from autograft harvesting from the knee and in part because of increasing availability and greater safety associated with allografts than previously achievable (Mroz et al., 2008; Savage-Elliott et al., 2014). The possibility exists that topographical variability, cartilage thickness, and mechanical properties between ankle and knee cartilage could be better matched if areas of the talus could be compared to areas of the distal femur. This would allow the surgeon an *á la carte* choice of femoral graft depending on the location of the talar OCL.

Cartilage exhibits a wide variety of depth- and location-dependent complex material behaviors. Shear and equilibrium cartilage moduli are substantially larger near the osteochondral interface than near the articular surface (Buckley et al., 2013, 2008; Chen et al., 2001; Wong and Sah, 2010; Wong et al., 2008a, 2008b), resulting in larger shear strains at the articular surface than in the bulk of the tissue (Wong and Sah, 2010; Wong et al., 2010, 2008a, 2008b). Further, most energy dissipation within cartilage occurs near the articular surface (Buckley et al., 2013, 2008). Spatial heterogeneity in depth-dependent cartilage mechanical properties have been reported in immature bovine knees (Silverberg et al., 2013), but such variations have not been investigated in human knees and ankles. Talar AOT grafts are commonly obtained from the patient's ipsilateral femoral condyle, although neither the biological nor the bulk mechanical properties match those of talar cartilage (Aurich et al., 2005; Fetter et al., 2006; Quinn et al., 2013; Schumacher et al., 2002; Treppo et al., 2000). To date, neither the depth-dependent mechanical behavior of native ankle cartilage nor that of potential graft tissue from unloaded regions of the knee has been well characterized. Therefore, the objective of this study was to characterize the depth-dependent shear properties and friction properties of human talar and donor-site femoral cartilage, and to establish the extent of variation in mechanical properties between these locations.

2. Materials and methods

2.1. Specimens

Nine tali and nine femurs were obtained from 14 cadaveric lower limbs (Table 1). Cadaveric tissue was obtained from an anatomic donation organization. During dissection, cartilage was screened for gross abnormality; macroscopically normal cartilage was used. Osteochondral cores were removed from nine talar anatomical regions (Elias et al., 2007) as well as from the superomedial and superolateral trochlear groove, common femoral donor-sites, using an 8 mm coring tool (Fig. 1A and B). Femoral locations were selected because of their common use as donor tissue in AOT procedures, in contrast to previous studies that have characterized mechanical behavior of weight-bearing regions (Wong and Sah, 2010; Wong et al., 2010, 2008a, 2008b). Samples were frozen at $-80\,°C$ between dissection and testing. Cylindrical cores were split into three samples: a 4 mm diameter osteochondral sample for tribology, a partial cylinder for shear, and the remainder for histology in a subset of samples (Fig. 1C).

Table 1
Specimen characteristics. An 'X' marks joints that were used in this study. Each ankle provided nine samples and each knee provided two samples.

Specimen	Gender	Age (years)	BMI (kg m^{-2})	Ankle		Knee	
				L	R	L	R
1	M	57	17		X		
2	F	47	23	X	X		
3	F	52	23	X			
4	F	47	24			X	X
5	M	69	19			X	
6	M	63	26				X
7	M	71	35			X	
8	M	75	35				X
9	M	60	25	X			
10	F	59	27		X		X
11	F	60	21	X		X	
12	M	48	18		X		
13	M	55	35	X			
14	M	49	30				X

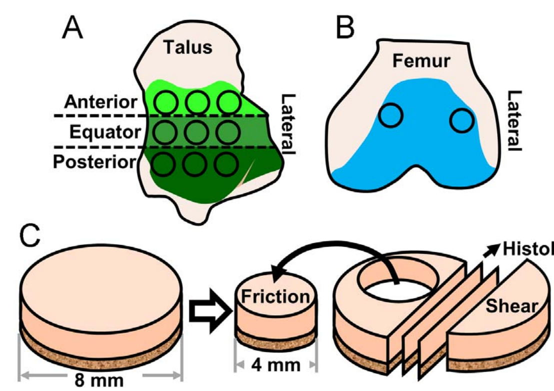

Fig. 1. (A) Osteochondral cores were removed from three anatomical regions on the talus, with three cores from each region. (B) Osteochondral cores were removed from two common donor-site locations on the femur. (C) Each sample was further divided into a 4 mm diameter sample for friction testing and a partial cylinder for shear testing. The remainder of a subset of samples was fixed and sectioned for histology.

2.2. Shear testing

Shear behavior was characterized using confocal elastography on a custom test device (Buckley et al., 2013, 2008) (Fig. 2A). Bone was trimmed to create a subchondral surface plane parallel to the articular surface, as confirmed visually under confocal microscopy (Buckley et al., 2008). Samples were stained in 28 μM 5-DTAF (Invitrogen, Waltham, MA) for 1–2 h. Following staining, samples were rinsed in PBS for 10 min to remove unbound dye. Samples were mounted on a custom backplate using cyanoacrylate. The moving front plate was sandblasted glass to provide a no-slip condition between the front plate and the cartilage following axial compression (Buckley et al., 2013). Cartilage thickness was measured under confocal microscopy; measurements were taken three times across the sample width and averaged. Samples were compressed axially and allowed to creep until reaching equilibrium, resulting in a final axial compression of $7.9 \pm 0.7\%$ (range, 6.8–9.7%). One to five lines were photobleached perpendicular to the articular surface (Fig. 2B). Cyclic shear loading was applied, using 1% shear strain at 1 Hz. While shear behavior varies with loading frequency (Buckley et al., 2013), 1 Hz was selected for its relevance to walking (Silverberg et al., 2013). Displacement of the photobleached lines was imaged during loading at 20 frames per second. After loading, the axial cross-sectional area was imaged for stress calculations.

Depth-dependent shear modulus, phase angle and energy dissipation were calculated using established methods (Buckley et al., 2013). Depth-dependent displacement of photobleached lines was tracked using custom code in MatLab (The MathWorks, Inc., Natick, MA), then used to calculate depth-dependent shear strain, shear modulus, phase angle and energy dissipation, by fitting displacement with a cosine function as described by Buckley et al. (2013).

Well-defined points on depth-dependent curves were obtained to compare between regions. Minimum and plateau shear moduli were evaluated (Fig. 2C). The minimum value was the global minimum; consistent with previous studies using different cartilage sources this typically occurred near the articular surface (Buckley et al., 2013, 2010; Silverberg et al., 2013). Plateau shear modulus was the average of