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Gait biomechanics in individuals with patellar tendon and hamstring tendon anterior cruciate ligament reconstruction grafts

Christopher D. Johnston [a,b], Jonathan S. Goodwin [a,b], Jeffery T. Spang [c], Brian Pietrosimone [a,b], J. Troy Blackburn [a,b,*]

[a] Department of Exercise and Sport Science, University of North Carolina at Chapel Hill, USA
[b] Program in Human Movement Science, University of North Carolina at Chapel Hill, USA
[c] Department of Orthopaedics, University of North Carolina at Chapel Hill, USA

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ABSTRACT

Anterior cruciate ligament reconstruction (ACLR) restores joint stability following ACL injury but does not attenuate the heightened risk of developing knee osteoarthritis. Additionally, patellar tendon (PT) grafts incur a greater risk of osteoarthritis compared to hamstring grafts (HT). Aberrant gait biomechanics, including greater loading rates (i.e. impulsive loading), are linked to the development of knee osteoarthritis. However, the role of graft selection on walking gait biomechanics linked to osteoarthritis is poorly understood, thus the purpose of this study was to compare walking gait biomechanics between individuals with HT and PT grafts. Ninety-eight (74 PT; 24 HT) subjects with a history of ACLR performed walking gait at a self-selected speed from which the peak vertical ground reaction force (vGRF) during the first 50% of the stance phase and its instantaneous loading rate, peak internal knee extension and valgus moments, and peak knee flexion and varus angles were obtained. When controlling for time since ACLR and quadriceps strength, there were no differences in any kinetic or kinematic variables between graft types. While not significant, 44% of the PT cohort were identified as impulsive loaders (displaying a heelstrike transient in the majority of walking trials) compared to only 25% of the HT cohort (odds ratio = 2.3). This more frequent observation of impulsive loading may contribute to the greater risk of osteoarthritis with PT grafts. Future research is necessary to determine if impulsive loading and small magnitude differences between graft types contribute to osteoarthritis risk when extrapolated over thousands of steps per day.

1. Introduction

The most common treatment for anterior cruciate ligament injury is surgical reconstruction (ACLR) using either patellar tendon (PT) or hamstring tendon (HT) autograft (Magnussen et al., 2010; Rahr-Wagner et al., 2014). ACLR restores joint stability (Keays et al., 2007; Sajovic et al., 2011), but does not attenuate the risk of knee osteoarthritis (Barenius et al., 2014; Luc et al., 2014). As many as 30–80% of patients develop osteoarthritis within 10–15 years (Barenius et al., 2014; Luc et al., 2014; Risberg et al., 2016), and negative changes in joint health are evident as early as 1–2 years post-ACLR (Culvenor et al., 2015; Eckstein et al., 2015).

Considerable research indicates that PT grafts incur a greater risk of osteoarthritis than HT grafts (Keays et al., 2007, 2010; Pinczewski et al., 2007; Poehling-Monaghan et al., 2017; Sajovic et al., 2011). However, the underlying causes of this discrepancy are unclear. Quadriceps dysfunction is a common long-term consequence of ACLR (Hart et al., 2010; Tengman et al., 2014), and has been implicated as a contributor to osteoarthritis development (Hart et al., 2011; Tourville et al., 2014). Greater quadriceps dysfunction has been reported with PT grafts compared to HT grafts (Keays et al., 2007; Mohtadi et al., 2011), potentially due to donor site morbidity, as a portion of its distal tendon is harvested for use as the graft. PT grafts also reportedly incur greater knee pain (Pinczewski et al., 2007; Poehling-Monaghan et al., 2017) which contributes to quadriceps dysfunction (Son et al., 2016).

Impulsive/high-rate loading results in greater cartilage degradation compared to lower loading rates in animal models (Ewers et al., 2002; Radin et al., 1991). In humans, higher loading rates during walking have been identified following ACLR (Blackburn

* Corresponding author at: University of North Carolina at Chapel Hill, 203 Fetzer Hall, CB# 8700, Chapel Hill, NC 27599-8700, USA.
E-mail address: troyb@email.unc.edu (J.T. Blackburn).

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et al., 2016a; Co et al., 1993; Noehren et al., 2013) and in individuals with knee osteoarthritis (Mundermann et al., 2005). The quadriceps attenuates loading in the early stance phase of gait, and quadriceps dysfunction results in impulsive loading (Blackburn et al., 2016b; Jefferson et al., 1990; Mikesky et al., 2000). Therefore, greater quadriceps dysfunction and impulsive loading during gait may contribute to the greater risk of osteoarthritis with PT grafts. However, impulsive loading characteristics have yet to be compared between individuals with HT and PT grafts.

Aberrant gait biomechanics including smaller knee flexion angles and sagittal knee moments, and larger ground reaction forces, knee varus angles, and frontal knee moments have been implicated in the mechanical pathogenesis of knee osteoarthritis (Chehab et al., 2014; Khandha et al., 2017; Kumar et al., 2014; Mundermann et al., 2005). Individuals with PT grafts reportedly display less knee flexion and smaller sagittal knee moments (Webster et al., 2004, 2005), and greater knee varus angles compared to those with HT grafts (Webster and Feller, 2011). While these findings suggest that PT grafts incur a gait biomechanics profile consistent with greater knee osteoarthritis risk, the sample sizes in these studies were relatively small (10–18 per group) and almost completely composed of males. Furthermore, these investigations exclusively evaluated early intervals post-ACLR (9–12 months) after which gait biomechanics reportedly change (Erhart-Hledik et al., 2018; Kaur et al., 2016). As such, the effects of graft type on gait biomechanics linked to knee osteoarthritis in individuals further removed from ACLR are unclear.

The purpose of this investigation was to compare walking gait biomechanics previously implicated in the mechanical pathogenesis of knee osteoarthritis between individuals with HT and PT grafts. We hypothesized that individuals with PT grafts would display greater impulsive loading and a biomechanics profile consistent with greater knee osteoarthritis risk (i.e. smaller knee flexion angles and internal knee extension moments; and larger ground reaction forces, knee varus angles, and internal knee valgus moments) compared to individuals with HT grafts.

2. Methods

2.1. Subjects

Power analysis of previous literature comparing gait biomechanics between PT and HT grafts (Webster and Feller, 2011, 2012; Webster et al., 2005) and impulsive loading characteristics between individuals with ACLR and healthy individuals (Co et al., 1993; Noehren et al., 2013) indicated that a sample of 7–37 subjects per graft cohort would provide power of 0.80 ($\alpha = 0.05$) to evaluate our hypotheses, with the exception of peak internal knee valgus moment for which 148 subjects per cohort would be necessary. A convenience sample of ninety-eight individuals with ACLR (74 PT; 24 HT) volunteered to participate in this cross-sectional study and were consecutively enrolled. Ninety-two of these individuals had undergone primary unilateral ACLR, while 6 subjects in the PT cohort had experienced multiple ACLR procedures (1 revision ACLR following graft rupture, 3 bilateral ACLR, and 2 bilateral ACLR and a revision ACLR). Subjects were at least 6 months post-ACLR and cleared by a physician to resume physical activity, and exercised regularly for at least 30 min 3× times per week. Subjects were excluded if they had a history of lower extremity injury within the 6 months prior to participation or a neurological disorder. The study was approved by the university's Institutional Review Board, and all subjects provided written informed consent prior to participation. Subject demographics are provided in Table 1.

2.2. Gait biomechanics assessment

Gait analysis was performed barefoot, and subjects wore spandex shorts and shirts or sports bras. We assessed gait biomechanics with subjects barefoot rather than shod to eliminate variability introduced by self-selected shoe type (e.g. control vs. neutral vs. cushioning) and condition (e.g. old vs. new), as well as the potential effects of laboratory-standardized shoes (e.g. neutral shoes in a subject who routinely wears pronation control shoes). Subjects walked at their preferred speed over embedded force plates (Bertec Corp. Columbus, OH) that were staggered so that the stance phase for both limbs could be assessed from a single trial. Subjects translated at least 3 m prior to heelstrike on the first force plate via 3–5 steps and completed at least 2 steps following heelstrike on the second force plate. Five practice trials were performed to determine the preferred walking speed with infrared timing gates. Five walking trials were then collected and retained for analysis if (1) each foot struck a separate force plate, (2) walking speed was within ±5% of the preferred speed during practice trials, and (3) gait kinematics were not visibly altered during the trial (e.g. "aiming" for the force plates).

Subjects were outfitted with 29 retroreflective markers (sternum and L4-L5 vertebral space, and bilateral acromioclavicular joints, anterior superior iliac spines, greater trochanters, anterior thighs, medial and lateral femoral epicondyles, anterior shanks, medial and lateral malleoli, first metatarsal heads, fifth metatarsal heads, and posterior calcanei) and a ridged cluster of three markers over the sacrum. A segment-linkage model of the lower extremities was constructed from a static calibration trial after which markers on the medial femoral epicondyles and malleoli were removed. Three-dimensional marker trajectories were sampled at 120 Hz using a 10-camera motion capture system (Vicon, Centennial, CO) and lowpass filtered at 10 Hz. Knee joint angles were calculated as motion of the shank segment relative to the thigh

Table 1
Subject demographics (mean ± sd).

	HT (n = 24)	PT (n = 74)	P-value
Sex[†]	19 females, 5 males	45 females, 29 males	0.101
Mass (kg)	67 ± 15	76 ± 19	**0.037**[*]
Height (m)	1.7 ± 0.1	1.7 ± 0.1	0.376
Age (years)	21 ± 4	22 ± 3	0.426
Time since ACLR (months)	43 ± 34	22 ± 26	**0.033**[*]
Gait speed (m/s)	1.3 ± 0.1	1.3 ± 0.1	0.859
Quadriceps strength (Nm/kg)	2.3 ± 0.6	1.8 ± 0.6	**0.001**[*]
KOOS pain	92 ± 6	90 ± 8	0.293
Tegner activity level	7.1 ± 1.8 (range 3–10)	6.5 ± 2.0 (range 3–10)	0.163

Bold text indicates statistical significance ($p < 0.05$).
[†] Compared between graft types via χ^2 analysis.
[*] Significantly different between graft types.

segment via Euler angles using a sagittal/frontal/transverse rotation sequence. Ground reaction forces were sampled at 1200 Hz and lowpass filtered at 75 Hz, and combined with kinematics and anthropometric data (Dempster et al., 1959) via standard inverse dynamics procedures (Gagnon and Gagnon, 1992) to yield net internal joint moments.

Analysis was restricted to the load acceptance phase (i.e. 1st 50% of stance) via custom software (LabVIEW, National Instruments Corp, Austin, TX). The stance phase was defined as the interval from heelstrike (vertical ground reaction force [vGRF] $\geq$ 20N) to toe off (vGRF $\leq$ 20N). The load acceptance phase is commonly assessed to evaluate how ground impact loading is attenuated throughout the lower extremity. This is particularly true with respect to impulsive loading, as the interval immediately following heelstrike displays the highest loading rate. Kinetic outcomes included the peak vGRF, peak instantaneous vGRF loading rate (first time derivative), and peak internal knee extension and valgus moments. Forces and loading rates were normalized to body weight (xBW and xBW$\cdot$s^{-1}, respectively) while moments were normalized to the product of body weight and height (xBW * Ht). Kinematic outcomes included the peak knee flexion and varus angles. Gait speed was calculated as the first time derivative of the sagittal plane coordinate of the pelvis center of mass and averaged over the interval from 0.5 m before and after heelstrike on the first force plate.

We also evaluated the presence of the heelstrike transient (HST), which is a rapid rise in the vGRF immediately following heelstrike that is indicative of impulsive loading (Hunt et al., 2010; Liikavainio et al., 2007). A 3-point sliding quadratic polynomial was fit to the vGRF, and changes in the sign of its 2nd derivative (i.e. changes in direction) were used to identify the temporal locations of vGRF "peaks". A trial was deemed to possess a HST if the ratio of the vGRF peak immediately following heelstrike to the subsequent local minimum exceeded 1.2 (Liikavainio et al., 2007; Radin et al., 1986). We recently demonstrated that this algorithm was the most effective of three different approaches for objectively identifying the heelstrike transient (Blackburn et al., 2016c).

2.3. Quadriceps function assessment

A subset of subjects completed isometric quadriceps strength assessment for the ACLR limb (PT = 57 [77% of cohort], HT = 18 [75% of cohort]). Subjects were seated on a dynamometer (HUMAC NORM, CSMi, Stoughton, MA) with the knee and hip in 90° and 85° of flexion, respectively, with straps secured over the torso, thigh, and leg, and were instructed to "kick out as hard and quickly as possible" against the dynamometer arm. Torque data were sampled at 600 Hz and lowpass filtered at 50 Hz, and quadriceps strength was defined as the largest peak torque value from 2 trials normalized to body mass.

2.4. Self-report pain

All subjects completed the Tegner Activity Level Scale to quantify their routine levels of physical activity. A subset of subjects completed the Knee Injury and Osteoarthritis Outcome Score (KOOS) survey (PT = 56 [76% of cohort], HT = 19 [79% of cohort]) from which the pain subscale was obtained to evaluate pain perception.

2.5. Statistical analysis

Normality of the outcomes was assessed via the Shaprio-Wilk test, visual inspection of histograms, and calculation of the ratio of skewness and kurtosis statistics to their standard errors. Outliers ($\geq$2.5 SD beyond the mean) were excluded from analyses to ensure normality. All analyses were performed using SPSS v21 statistical software (IBM Corp., Armonk, NY), and significance was established *a priori* as α = 0.05.

Gait biomechanics outcomes were evaluated via mixed-model 2 (graft type: PT vs. HT) $\times$ 2(limb: ACLR vs. contralateral) repeated-measures ANOVA. Graft type $\times$ limb interaction effects and limb main effects are addressed in the Results. Aberrant gait biomechanics have been reported in the ACLR limb relative to the contralateral limb up to 5 years post-ACLR (Kaur et al., 2016), while others have reported symmetry between limbs (Blackburn et al., 2016a; Khandha et al., 2017). Therefore, while the analysis plan primarily focused on comparisons of the ACLR limb between the PT and HT cohorts, we also compared gait biomechanics between limbs to determine if aberrant gait biomechanics were present in our sample.

Gait speed influences biomechanical outcomes. Similarly, changes in gait biomechanics occur across the spectrum of time since ACLR represented by our sample (range = 6–161 months) (Erhart-Hledik et al., 2018; Kaur et al., 2016). The level of routine physical activity following ACLR may also influence gait biomechanics. Gait speed, time since ACLR, quadriceps strength, Tegner Activity Level, and the KOOS pain subscale were compared between groups via independent t-tests. The ANOVA models described above were then repeated using characteristics that differed between graft cohorts as covariates (i.e. 2 $\times$ 2 ANCOVA).

Subjects were classified as Impulsive Loaders (HST identified in at least 3 of 5 trials) or Normal Loaders (HST identified in 2 or fewer trials) based on whether the majority of gait trials displayed the HST in the ACLR limb (Blackburn et al., 2016c). χ^2 analysis (2-way crosstabs) was conducted to compare the frequencies of Impulsive vs. Normal loaders across graft cohorts. We also calculated the relative risk and odds ratio for being identified as an Impulsive Loader in the respective cohorts.

Because 6 individuals in the PT cohort experienced multiple ACLR procedures (e.g. revision following graft rupture and/or bilateral ACLR), we repeated the analyses described above after excluding these subjects.

3. Results

All outcomes were normally distributed following removal of outliers. Sample sizes for each analysis after removing outliers are indicated in parentheses below. Gait speed was identical across the groups (p = 0.859), and KOOS pain scores (p = 0.293) and Tegner Activity Level scores (p = 0.163) were highly similar (Table 1). However, Time Since ACLR (p = 0.033) and Quadriceps Strength (p = 0.001) were greater in the HT cohort (Table 1), and were used as covariates in subsequent analyses. Removing the 6 subjects with multiple ACLR procedures generally did not alter the results with the exception of 2 minor findings indicated below.

3.1. Kinetics

Peak vGRF did not differ between the ACLR and Contralateral limbs in the primary analysis (PT = 71, HT = 24; p = 0.084), but was greater in the Contralateral limb (PT = 55, HT = 18; p = 0.006) after covarying for Time Since ACLR and Quadriceps Strength, and this difference persisted after removing the 6 individuals in the PT cohort with multiple ACLR procedures. Conversely, no differences were identified between graft types (Table 2). No differences were present between limbs or graft types for the vGRF loading rate (PT = 71, HT = 24; p = 0.453).

Peak internal knee extension moment was greater in the Contralateral limb (PT = 72, HT = 22; p = 0.020), and this difference

Table 2
Gait kinetics (mean ± sd).

	ACLR	Contralateral
Peak vGRF (xBW)		
HT	1.12 ± 0.08	1.12 ± 0.09
PT	1.07 ± 0.07	1.10 ± 0.09
Collapsed across graft types*	**1.08 ± 0.08**	**1.11 ± 0.09**
vGRF loading rate ($xBW{\cdot}s^{-1}$)		
HT	53.35 ± 14.44	55.15 ± 14.69
PT	52.40 ± 15.73	52.73 ± 15.22
Collapsed across graft types	52.87 ± 18.04	53.94 ± 17.65
Peak internal knee extension moment (xBW * Ht)		
HT	−0.036 ± 0.014	−0.038 ± 0.012
PT	−0.032 ± 0.016	−0.039 ± 0.016
Collapsed across graft types†	**−0.035 ± 0.019**	**−0.038 ± 0.019**
Peak internal knee valgus moment (xBW * Ht)		
HT	−0.025 ± 0.007	−0.027 ± 0.004
PT	−0.026 ± 0.007	−0.027 ± 0.008
Collapsed across graft types	−0.025 ± 0.010	−0.027 ± 0.010

Bold text indicates statistical significance ($p < 0.05$).
Negative (−) values reflect extension and valgus.
* Greater in Contralateral limb after covarying for the combined effect of Quadriceps Strength and Time Since ACLR (p = 0.006).
† Greater in the Contralateral limb in the primary analysis (p = 0.048) and after covarying for the combined effect of Time Since ACLR and Quadriceps Strength (p = 0.039).

Table 3
Contingency table for Heelstrike transient [# of subjects (% of cohort)].

	PT	HT
Impulsive loaders	32 (44%)	6 (25%)
Normal loaders	41 (56%)	18 (75%)
Relative risk of being an impulsive loader (95% CI)	1.730 (0.825, 3.267)	
Odds ratio of being an impulsive loader (95% CI)	2.286 (0.814, 6.146)	

χ^2 analyses indicated that the frequencies of Impulsive vs. Normal Loaders did not differ between graft cohorts (p = 0.148). The Relative Risk and Odds Ratio were calculated as the odds/risk in the PT cohort relative to the HT cohort. These ratios were not statistically significant, as the 95% confidence intervals crossed 1.

persisted after covarying for Time Since ACLR and Quadriceps Strength (PT = 56, HT = 15; p = 0.035). However, no differences were identified between graft types (Table 2). Removing subjects with multiple ACLR procedures did not change the overall analysis; however, the limb main effect was no longer present after covarying for Time Since ACLR and Quadriceps Strength (ACLR = −0.035 ± 0.014 xBW * Ht, CON = −0.040 ± 0.014 xBW * Ht; p = 0.073). No differences were present between limbs or graft types for the peak internal knee valgus moment (PT = 70, HT = 23; p = 0.697).

A contingency table detailing the frequencies of Impulsive and Normal loaders in each graft cohort is presented in Table 3. The percentage of subjects identified as Impulsive Loaders was greater in the PT cohort than in the HT cohort (44% vs. 25%), but did not differ statistically between graft types (p = 0.148). Similarly, the odds ratio for being an Impulsive Loader in the PT cohort relative to the HT cohort was 2.286 (95% CI: 0.814, 6.146) and the relative risk of being an Impulsive Loader in the PT cohort was 1.730 (95% CI: 0.825, 3.267). Though the odds and risk of being identified as an Impulsive Loader were greater in the PT cohort as indicated by the high upper limit of the 95% CI, these ratios were not statistically significant.

Table 4
Gait kinematics (mean ± sd).

	ACLR	Contralateral
Peak knee flexion angle (°)		
HT	9.7 ± 5.1	10.6 ± 4.0
PT	8.5 ± 6.0	10.1 ± 6.0
Collapsed across graft types*	**9.1 ± 6.8**	**10.4 ± 6.5**
Peak knee varus angle (°)		
HT	−0.8 ± 2.9	−0.5 ± 2.7
PT	0.6 ± 3.2	0.2 ± 3.0
Collapsed across graft types	−0.1 ± 3.6	0.1 ± 3.4

Bold text indicates statistical significance ($p < 0.05$).
Negative (−) values reflect extension and valgus.
* Greater in the Contralateral limb in the primary analysis (p = 0.022) and after covarying for the combined effect of Time Since ACLR and Quadriceps Strength (p = 0.009).

3.2. Kinematics

Peak knee flexion angle was smaller in the ACLR limb (PT = 73, HT = 23; p = 0.022), and this difference persisted after controlling for Time Since ACLR and Quadriceps Strength (PT = 55, HT = 18; p = 0.009), and after removing individuals with multiple ACLR procedures. However, no differences were observed between graft types (Table 4). No differences were present between limbs or graft types for the peak varus angle (PT = 71, HT = 24; p = 0.248).

4. Discussion

This is the most comprehensive cross-sectional comparison of gait biomechanics between ACLR graft types, to our knowledge, as it evaluated both joint kinematics and kinetics and impulsive loading characteristics linked to cartilage degradation in a relatively large sample compared to similar previous investigations (Webster and Feller, 2011; Webster et al., 2004, 2005). Our data indicate that while gait abnormalities that could negatively influence joint health are present in the ACLR limb (e.g. smaller ground reaction forces and knee extension moments) (Khandha et al., 2017; Pietrosimone et al., 2017; Wellsandt et al., 2016), gait biomechanics did not differ between graft cohorts.

Our results agree with previous work reporting no differences in peak knee flexion angle, vGRF, or frontal plane moments between HT and PT grafts (Webster and Feller, 2012; Webster et al., 2005). Contrary to our results, Webster and Feller (2011) reported that peak knee varus angle was greater in individuals with PT vs. HT. However, a subsequent study by this group found that peak varus angle did not differ between graft cohorts (Webster and Feller, 2012), similar to our results. This group also reported smaller external knee flexion moments in individuals with PT grafts (Webster et al., 2005) which is in contrast to our finding of no difference in internal knee extension moment. Both graft cohorts studied by Webster et al. were composed of 94% males compared to 39% and 21% in our PT an HT cohorts, respectively. As sex-specific differences in sagittal plane moments have been reported post-ACLR (Di Stasi et al., 2015), the differences in sex distributions across studies may have influenced the outcomes. Our study was not adequately powered to account for the influence of sex, but future research should evaluate the combined influence of graft type and sex on gait biomechanics post-ACLR.

Animal models indicate that higher loading rates (i.e. impulsive loading) result in greater disruptions of cartilage structure and biosynthesis compared to lower-rate loading (Chen et al., 1999; Ewers et al., 2002; Kurz et al., 2001; Radin et al., 1991; Yang et al., 1989). We evaluated impulsive loading via the instantaneous vGRF loading rate and presence of the HST. Though no statistically significant differences in these characteristics were noted between graft cohorts, 44% of the PT cohort were identified as Impulsive loaders (i.e. HST in the majority gait trials) compared to only 25% in the HT cohort, resulting in a 1.73× greater relative risk of individuals with PT grafts being identified as Impulsive loaders. Though this relative risk was not statistically significant, the upper bound of the 95% CI (0.825, 3.267) suggests that the risk of being an Impulsive Loader may be more than 3× greater in individuals with PT. More frequent observation of the HST may contribute to the greater risk of osteoarthritis with PT grafts when extrapolated over thousands of steps per day. However, future research is necessary to determine the long-term implications of this loading characteristic for joint health.

Quadriceps dysfunction (e.g. weakness, atrophy, central activation failure, etc.) is common following ACLR (Hart et al., 2010; Tengman et al., 2014), and contributes to aberrant gait biomechanics and the heightened risk of osteoarthritis following ACLR (Blackburn et al., 2016b; Hart et al., 2011; Segal et al., 2010; Tourville et al., 2014). One of the quadriceps' fundamental roles is to attenuate impact forces during gait, and quadriceps dysfunction is associated with greater loading rates (Blackburn et al., 2016b; Jefferson et al., 1990; Mikesky et al., 2000). Similar to previous long-term follow-ups (Keays et al., 2007; Mohtadi et al., 2011), the quadriceps was significantly weaker in our PT cohort. However, covarying for the influence of quadriceps strength did not alter comparisons of gait biomechanics between the graft cohorts. Similarly, time since ACLR was notably greater in the HT cohort, yet gait biomechanics did not differ between graft cohorts either when this influence was present or was accounted for statistically.

Though the KOOS pain subscale was similar in our PT and HT cohorts, multiple longitudinal investigations and meta-analyses have reported a greater incidence of pain with PT grafts (Freedman et al., 2003; Mohtadi et al., 2011; Pinczewski et al., 2007; Poehling-Monaghan et al., 2017). This discrepancy is potentially due to the fixed follow-up intervals in these previous investigations vs. the wide range of time since ACLR in our sample (6–161 months). The greater incidence of knee pain with PT grafts reported in these longitudinal follow-ups could manifest as a lower level of physical activity (Mascarenhas et al., 2012; Pinczewski et al., 2007; Rahr-Wagner et al., 2014), creating an environment that promotes obesity, inflammation, altered joint loading, and poor joint health. Additionally, a higher incidence of meniscal and chondral injuries has been reported with PT grafts (Keays et al., 2007; Poehling-Monaghan et al., 2017), which likely contributes to the greater osteoarthritis risk (Barenius et al., 2014; Keays et al., 2010; Ruano et al., 2016).

Interpretation of the results of this study should be considered within the context of its limitations. The risk of developing osteoarthritis is higher in females compared to males (Thomas et al., 2016; Zhang and Jordan, 2010), and sex-specific differences in gait biomechanics have been reported post-ACLR (Asaeda et al., 2017; Di Stasi et al., 2015; Webster et al., 2012). We combined males and females in our analyses in an effort to maximize the sizes of the HT and PT cohorts, potentially masking sex differences that may influence osteoarthritis risk. Future studies should continue to investigate the effects of sex on walking gait biomechanics after ACLR and the influence on osteoarthritis risk, and should incorporate more balanced sample sizes given the relatively small size of our HT cohort. Our analysis plan was also somewhat conservative, as removing statistical outliers may exclude individuals who provide important information. However, only 1–4 subjects were removed from any given analysis, and the outliers did not generally constitute a distinct group (i.e. different subjects were excluded from different analyses), suggesting that the outliers were random in nature and unlikely to provide additional insight. Additionally, the cross-sectional nature of the study precludes our ability to determine the long-term implications of ACLR graft type and gait biomechanics on osteoarthritis development. Other outcomes such as joint contact forces (Saxby et al., 2016) may be warranted in future research. Similarly, high levels of quadriceps/hamstrings co-contraction during gait result in greater joint compressive force (Tsai et al., 2012), and longer duration co-activation is associated with faster progression of knee OA (Hodges et al., 2016). Incorporating EMG-driven models to estimate knee joint contact forces may elucidate additional insight.

In conclusion, the results of this study do not support our hypothesis that gait biomechanics linked to knee osteoarthritis development following ACLR differ between individuals who have received PT and HT grafts. Future prospective longitudinal research is necessary to identify characteristics that influence the greater risk of osteoarthritis in individuals who receive PT grafts.

Conflict of interest statement

None of the authors has a financial or personal relationship with other people or organizations that could inappropriately bias his/her work.

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