Quadriceps strength is not related to gait impact loading in knee osteoarthritis

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ABSTRACT

Joint loading has been implicated in the pathogenesis of knee osteoarthritis (OA). While compartment-specific measures such as the knee adduction moment have received much attention in the literature, less is known about other measures of dynamic loading in this patient population. This cross-sectional study assessed strength and walking patterns of 204 individuals with radiographically confirmed medial tibiofemoral OA and varus malalignment. Pearson product moment correlations and regression analyses were used to determine the bivariate and multivariate relationships amongst measures of impact loading (rate of loading and heelstrike transient occurrence) with demographic, clinical (in particular, radiographic disease severity, lower limb alignment, and self-reported pain and function), and biomechanical variables (maximum voluntary isometric quadriceps strength and gait kinematics). While maximum voluntary isometric quadriceps strength was significantly correlated with rate of loading ($r=0.27$) when walking at a freely chosen speed, multiple regression analyses indicated that rate of loading was primarily dictated by walking speed ($p<0.001$), and the effect of quadriceps strength was insignificant when accounting for all other included variables. Individuals who exhibited a heelstrike transient in their vertical ground reaction force profile were significantly more varus malaligned and were more likely to demonstrate severe radiographic degeneration than those who did not exhibit heelstrike transients. These results demonstrate higher impact loading during walking in those with knee OA with faster self-selected walking speeds, though the relationship with quadriceps strength is less clear. Importantly a potential association between disease characteristics, such as malalignment and disease severity, and higher impact loading was also observed.

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1. Introduction

Walking is the most common activity of daily living and is associated with the largest overall cumulative load at the knee — the weight-bearing joint most commonly affected by osteoarthritis (OA) [1]. As such, walking has provided a useful paradigm for examining the role of loading in the pathogenesis of knee OA in humans [2–4]. However, the direct measurement of knee joint load during walking in vivo is difficult for many reasons and has necessitated the development of proxy measures for this construct. Knee joint loading in a variety of ways. The external knee adduction moment (KAM), has received much attention in the OA literature due to its association with clinical outcomes such as disease severity [5], and progression [6], bone mineral density [7], and symptoms such as pain [8]. However, the KAM is a compartment-specific measure of load with maximum values occurring at approximately 30 and 70% of stance [9]. In contrast, high impact loading is known to occur immediately following initial contact and is not constrained to a single compartment of the knee [10,11]. Thus, it may provide important additional information regarding the anatomical and biomechanical outcomes observed in those with knee OA.

Although some studies have described discrete impact loading characteristics during walking in healthy individuals [10–13], very few pertain to those with knee OA [14,15]. Impact loading characteristics have typically been quantified based on the presence or absence of a heelstrike transient (HST) in the ground reaction force (GRF) profile as well as rates of change in the GRF magnitude (i.e. rate of loading). In addition to high rates of loading, some individuals also exhibit a HST within the first 50 ms of stance — a distinctive, sharp peak in the GRF curve suggested to occur in one-third of adults [16]. Since high rates of loading and HST presence do not necessarily occur concurrently, both of these outcome measures have been used independently when quantifying impact loading during walking. Indeed, it is the potentially damaging impact loading experienced immediately following initial contact that has led some authors to implicate it as an important contributor to OA development and
Disease severity was assessed by the same two examiners using the weight-bearing short progression [17–20]. As a result, understanding factors which influence the magnitude and frequency of these impact loading characteristics represents an important framework for preventing and treating knee OA.

Increased activation of the quadriceps immediately preceding initial contact can reduce the potential for HST occurrence [15], while experimentally-induced quadriceps paralysis can increase HST magnitude in healthy individuals [21]. Additionally, a cross-sectional comparison of sedentary women with those who participated in strength training [10] showed higher rates of loading and more frequent HST occurrence in the sedentary group. The sedentary group also exhibited lower isokinetic quadriceps strength relative to body mass. Although these findings point to a potential influence of quadriceps muscle strength on impact loading during walking, no studies have examined this relationship in knee OA. This is of particular importance given that quadriceps muscle weakness is common in knee OA [22–24], reduced quadriceps strength relative to body mass is predictive of disease development [22], low absolute quadriceps strength predicts poor functional outcome [25], and quadriceps strength decreases have been linked to loss of tibiofemoral joint space [26].

The primary purpose of this study was to investigate measures of impact loading during walking (rate of loading and HST occurrence) in individuals with knee OA and their relationship with quadriceps muscle strength. A secondary purpose was to examine demographic, clinical and biomechanical differences between those who exhibit high rates of impact loading as well as HSTs and those who do not.

2. Materials and methods

2.1. Participants

The present study provides data from individuals with knee OA recruited for two separate, but similar, cohorts. Inclusion and exclusion criteria and recruitment strategies were the same for both cohorts and, with the exception of radiographic analysis, all data were collected and analyzed using identical techniques. Data from 204 individuals were used in the study (91 M, 113 F; mean age = 64.7, sd 8.6 years; mean height = 1.66, sd 0.09 m; mean mass = 80.6, sd 15.0 kg). In cases of bilateral symptoms, the eligible knee reported to be most painful was denoted the study knee. Based on the Kellgren and Lawrence (KL) grading system [27], seventy-seven individuals had mild (KL grade 2), 67 had moderate (KL grade 3), and 60 had severe (KL grade 4) radiographic OA. All participants had self-reported medial knee pain during walking >3 on an 11-point scale (0 = no pain; 10 = maximal pain). Participants were excluded if they had a history of lower limb joint replacement; knee surgery (including arthroscopy) within the previous 6 months; a diagnosis of rheumatoid or other systemic arthritis; were currently receiving physiotherapy for knee OA; had a severe medical condition that affected gait; or had valgus lower limb alignment based on radiographic analysis (see below). Initial screening was conducted via telephone interviews and those who met study criteria underwent radiographic examination to confirm a diagnosis of medial tibiofemoral OA. Ethical approval was obtained from the University of Melbourne Human Research Ethics Committee. All participants provided written informed consent prior to any assessment.

2.2. Radiographic analysis

Over the course of data collection, different radiographic protocols were used for the two cohorts. One hundred and forty five participants underwent a single anteroposterior, weight-bearing short film X-ray in full knee extension, while the remainder (n = 59) of study participants underwent a single, semi-flexed, posteroanterior weight-bearing short film X-ray with the feet externally rotated 10°. Disease severity was assessed by the same two examiners using the Kellgren and Lawrence grading system [27]. In cases where there was disagreement between examiners, a third examiner was brought in and consensus was reached amongst the examiners. Mechanical lower limb alignment for those undergoing the semi-flexed X-ray (n = 59) was measured on the study limb using techniques and equations described by Kraus et al. [28], while alignment for the remainder (n = 145) was measured using methods described by Moreland et al. [29] and regression equations for mechanical alignment provided by Hinman et al. [30].

2.3. Self-reported symptoms and function

Participants completed the Likert version of the Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC) [31]. Subscale scores were calculated (maximum scores: pain = 20, stiffness = 8, function = 68), with higher scores indicating more pain, stiffness and physical dysfunction.

2.4. Quadriceps strength

Isometric quadriceps muscle strength was assessed at 60° of knee flexion using the Kin-Com 125-AP dynamometer (Chattecx Corporation, Chattanooga, TN, USA). Participants were seated in a standardized position of 90° of hip flexion and stabilized with torso and waist straps as well as a thigh block. The lever arm axis of rotation was aligned with the lateral femoral condyle for each patient. Participants performed a single, submaximal trial to ensure familiarization with the protocol and testing apparatus. Participants were then instructed to perform three trials of maximal effort isometric knee extension lasting 5 s in duration, separated by 15 s of rest. The maximum force output from the 3 trials was recorded and converted to Nm by multiplying by the lever arm distance and then normalized to body mass (Nm/kg) to account for the effect of mass on torque development.

2.5. Gait analysis

Participants underwent a single, three-dimensional gait analysis. Kinematic data were collected at 120 Hz using a six-camera Vicon motion analysis system (Vicon, Oxford, UK). Passive-reflective markers were affixed to the skin according to the standard Plug-In Gait marker set. Specifically, markers were placed bilaterally over the following landmarks: anterior and posterior superior iliac spines, lateral aspect of the thigh, lateral femoral epicondyle, lateral aspect of the shank, lateral malleolus, calcaneus, and top of the foot at the base of the 2nd metatarsal. Medial knee and ankle markers were included during an initial static standing trial to determine relative positioning of joint centres, and were then removed prior to gait testing. Kinetic data were collected using two, floor-mounted force plates at a sampling rate of 1080 Hz (Advanced Mechanical Technology Inc., Watertown, MA), in synchrony with the cameras.

Participants were instructed to walk in their own low-heeled sneakers at a self-selected walking speed that was measured using photoelectric timing gates (Jaycar Electronics, Melbourne, Australia), attached to a stopwatch, and placed along the walkway 2 m on either side of the force plates. A total of five trials with clean, single force plate strikes from the study limb were collected. Values for walking speed and for each gait variable listed below were obtained by averaging across the trials.

The magnitudes of knee flexion at the instant of initial contact (defined as the point where the vertical ground reaction force (GRF) exceeded 20 N) as well as the maximum amount of knee flexion occurring during loading were identified for each trial. Instantaneous vertical ankle velocity at initial contact was calculated as the first derivative of the ankle marker vertical displacement data. Additionally, rates of loading were calculated from the vertical component of the ground reaction force (GRFz) using a computer-driven algorithm.
with all values normalized to body mass (expressed in body weight (BW) units):

i) \( F_z \) peak 1 magnitude (BW) — defined as the maximum value of the first \( GRF_z \) (i.e. vertical component) peak. This value corresponds to the magnitude of the largest vertical force exerted to the body during weight acceptance. The point of stance phase where this occurred was also identified.

ii) Average loading rate (BW/s) — defined as the \( F_z \) peak magnitude divided by the time taken from initial contact to that point. This gives an indication of average impact loading experienced by the lower limb throughout weight acceptance.

iii) Max loading rate (BW/s) — defined as the maximum rate of change of the \( GRF_z \), determined by calculating the first order derivative of force versus time (i.e. slope) and identifying the maximum rate. This value denotes the single largest immediate change in vertical force experienced during weight acceptance. The point in stance where this occurred was also identified.

A clear classification system for identifying HSTs is not available in the literature, as no clear distinction between them has been made on biomechanical grounds. We operationally defined HSTs based on the following rationale: given the consistent finding of higher transient spikes occurring later in the \( GRF \) upslope, we chose to concentrate on clearly evident HSTs in the upper 50% of the participant’s \( F_z \) peak 1 magnitude. A HST was determined to have occurred if, during upper 50% of the \( GRF \), upslope (i.e. between 50% of the \( F_z \) peak 1 magnitude and the actual \( F_z \) peak 1 magnitude), the magnitude of the \( GRF \) peaked and then decreased by more than 0.5% of the \( F_z \) peak 1 magnitude (see Fig. 1). To ensure totally divergent groups for secondary analyses (i.e. to examine differences between those who exhibit HSTs and those who do not), an individual was deemed to exhibit a HST if a HST was observed in greater than 75% of their trials while an individual was deemed to not exhibit a HST if a HST was not observed in any of their trials (see Fig. 2).

2.6. Statistical analysis

The Statistical Package for the Social Sciences v16 (SPSS Inc., Chicago, USA) was used for all data analyses. Pearson product moment correlations were used to examine bivariate relationships between normalized maximum isometric quadriceps torque with walking speed and the three aspects of the \( F_z \) curve described above. Associations amongst variables were explored further using forced entry multiple linear regression to determine the amount of variance in the three \( F_z \) curve variables explained by the following predictor variables: KL grade, alignment, quadriceps torque, WOMAC pain and function, walking speed, and ankle velocity and knee flexion angle at initial contact.

Demographic, clinical, and biomechanical variables were all compared between HST groups. With the exception of radiographic severity, independent sample \( t \)-tests were used. Differences in radiographic severity were examined using a chi-square test. Lastly, to predict the HST occurrence based on the predictor variables described above, a forced entry multiple logistic regression model was used. Given the multiple analyses used, a conservative alpha level of 0.01 was set to denote significance for all statistical tests.

3. Results

Participants, on average, produced \( 1.32 \pm 0.53 \) N m/kg of maximal isometric quadriceps torque. Group data for gait variables are summarized in Table 1. Significant relationships existed between quadriceps torque and rate of loading variables at the freely chosen walking speed (\( r > 0.27; p < 0.01 \)). A stronger quadriceps muscle tended to be associated with walking faster and higher loading magnitudes and rates of loading.
Regression analyses indicated that the clinical and biomechanical variables examined could explain 59% of the variance in the F_T peak 1 magnitude, 49% of the variance in the average F_T loading rate, and 38% of the variance in maximum F_T loading rate (Table 2). Freely chosen walking speed was a significant contributor in two of the regression models (maximum F_T loading rate and F_T peak 1 magnitude), maximum knee flexion at initial contact was significant in the same two models, and KL grade was a significant contributor in explaining variance in F_T peak 1 magnitude. No other independent variables were significant contributors to any regression model (p > 0.01).

Clinical and biomechanical data comparing those who exhibited HSTs and those who did not are presented in Table 3. Thirty-five (17%) individuals did not exhibit a HST in any walking trials; thirty-nine (19%) exhibited a HST in greater than 75% of trials; while one-hundred and thirty (64%) participants exhibited a HST in at least one walking trial, but not in more than 75%. The HST group was significantly more varus than the group who did not exhibit HSTs (p < 0.01). There was also a trend for more participants with severe radiographic changes (KL grade 4) to be in the HST group, while the non-HST group tended to have more participants with mild radiographic changes (KL grade 2) (χ² = 6.53, p = 0.04). Compared to the non-HST group, those in the HST group also exhibited significantly greater maximum loading rates (p = 0.002) and a trend towards earlier times to maximum loading rate and lower average loading rates. No other between-group differences were found when comparing demographic, clinical, or biomechanical variables.

Logistic regression identified which clinical and biomechanical variables predicted the presence of a HST. The combination of WOMAC pain and function, KL grade, alignment, quadriceps torque, walking speed, and knee flexion and ankle velocity at initial contact was able to correctly predict the classification (HST or non-HST) of 52 (70.3%) participants. Despite this, alignment was the only predictor variable to approach significance (p = 0.05) in the model.

### 4. Discussion

This study is the first to examine the relationship between quadriceps strength and rate of loading during walking in individuals with knee OA. Results indicate that stronger quadriceps are associated with a faster walking speed and higher rates of impact loading. However, the relationship between quadriceps strength and rate of loading became non-significant when controlling for other variables (including walking speed) via multiple regression. This is also the first study to distinguish between people who exhibit HSTs from those who do not in this patient population, with those exhibiting HSTs having more varus malalignment. Results from this study add to the growing body of evidence implicating joint loading in the pathogenesis of knee OA and its association with disease characteristics.

The role of quadriceps strength in the attenuation of loading and its subsequent effects on knee OA pathogenesis has received much interest in the literature. Although many authors have suggested a theoretical basis for this role, little data exists in this patient population to provide direct support. Instead, studies investigating the effects of experimentally-induced quadriceps paralysis [21], knee pain [20], or altered neuromuscular activation patterns [13] in healthy individuals point to the need for sufficient quadriceps activation prior to initial contact to reduce impact loading and/or the presence of HSTs. In the only study to date reporting a relationship between quadriceps strength and rate of loading during walking, Mikesky et al. [10] showed that healthy women actively engaged in strength training exhibit lower normalized maximum loading rates (BW/s) and fewer HSTs when walking at a fixed velocity (1.22–1.35 m/s) than a group of sedentary women; although a direct correlation between quadriceps strength and rate of loading was not provided.

Although maximum voluntary quadriceps torque in the present study was significantly associated with loading rate magnitudes when participants walked at their freely chosen walking speed (r = 0.27, p = 0.008), it is likely that walking speed mediated much of the relationship between quadriceps strength and loading rates. Specifically, those with stronger quadriceps walked faster and subsequently experienced higher rates of loading. This hypothesis is supported by results of the regression analyses and a significant correlation between quadriceps strength and walking speed. When controlling for other clinical and biomechanical variables using multiple linear regression, quadriceps torque was unable to explain a significant independent portion of variance in loading rate, despite having sufficient statistical power. Importantly, it is not known whether stronger quadriceps permit faster walking with associated higher impact loading or conversely, if faster walking results in stronger quadriceps, again with the theoretical basis for this role, little data exists in this patient population to provide direct support. Instead, studies investigating the effects of experimentally-induced quadriceps paralysis [21], knee pain [20], or altered neuromuscular activation patterns [13] in healthy individuals point to the need for sufficient quadriceps activation prior to initial contact to reduce impact loading and/or the presence of HSTs. In the only study to date reporting a relationship between quadriceps strength and rate of loading during walking, Mikesky et al. [10] showed that healthy women actively engaged in strength training exhibit lower normalized maximum loading rates (BW/s) and fewer HSTs when walking at a fixed velocity (1.22–1.35 m/s) than a group of sedentary women; although a direct correlation between quadriceps strength and rate of loading was not provided.

### Table 1

<table>
<thead>
<tr>
<th>Variable</th>
<th>F_T peak 1 magnitude</th>
<th>Average F_T loading rate</th>
<th>Maximum F_T loading rate</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>β-coefficient</td>
<td>p-value</td>
<td>β-coefficient</td>
</tr>
<tr>
<td>Model: (R² = 0.59, p &lt; 0.001)</td>
<td></td>
<td></td>
<td>Model: (R² = 0.49, p &lt; 0.001)</td>
</tr>
<tr>
<td>F_T peak 1 magnitude</td>
<td>0.75 (0.67, 0.83)</td>
<td>&lt;0.001</td>
<td>-13.93 (-33.50, -4.38)</td>
</tr>
<tr>
<td>β0</td>
<td>-0.02 (-0.03, -0.01)</td>
<td>0.01</td>
<td>0.07 (-1.97, 0.77)</td>
</tr>
<tr>
<td>Lower limb alignment</td>
<td>-0.01 (-0.01, 0.01)</td>
<td>0.69</td>
<td>0.05 (-0.80, 0.21)</td>
</tr>
<tr>
<td>WOMAC pain</td>
<td>-0.01 (-0.01, 0.01)</td>
<td>0.98</td>
<td>-0.27 (-0.80, 0.21)</td>
</tr>
<tr>
<td>WOMAC function</td>
<td>0.01 (0.01, 0.01)</td>
<td>0.57</td>
<td>0.06 (-0.07, 0.22)</td>
</tr>
<tr>
<td>Quadriceps torque</td>
<td>0.01 (0.01, 0.01)</td>
<td>0.17</td>
<td>1.29 (-0.11, 4.27)</td>
</tr>
<tr>
<td>Walking speed</td>
<td>0.30 (0.24, 0.36)</td>
<td>&lt;0.001</td>
<td>25.63 (20.71, 34.52)</td>
</tr>
<tr>
<td>Knee flexion at initial contact</td>
<td>0.01 (0.01, 0.01)</td>
<td>&lt;0.001</td>
<td>0.23 (0.15, 0.31)</td>
</tr>
<tr>
<td>Ankle velocity at initial contact</td>
<td>0.01 (0.01, 0.01)</td>
<td>0.37</td>
<td>-0.09 (-1.27, 1.56)</td>
</tr>
</tbody>
</table>
be found from previous studies investigating the effects of quadriceps strengthening interventions on the KAM during walking in those with knee OA. Though few studies exist, they have all found a trend toward increases in the KAM after a strengthening program [32–34]. These findings, together with ours, suggest that the role of quadriceps strength on different measures of loading is unclear and may be explained by improved functioning associated with stronger muscles. Clearly, further research in this area is needed.

Importantly, our method of quantifying maximum quadriceps strength may have played a role in our findings. We chose to measure isometric knee extensor strength in 60° of knee flexion for a couple of reasons. This angle corresponds to the region where the knee extensor mechanism generates its greatest torque [35] and measurement of strength at smaller flexion angles may increase variability due to the position on the torque-joint angle curve (i.e. on the ascending arm, small changes in joint angle could potentially result in large changes in strength output). Also, the angular velocity of the knee around initial contact during walking is relatively low, and the slope of the torque-velocity relationship in the slow-velocity-high torque region is relatively flat [23,36]. Isometric muscle testing also provides a reliable and consistent means to assess maximal muscle strength. That said, the limitations of measuring isometric strength to imply muscle function during dynamic activities are well known. Further, humans do not activate fully their quadriceps during walking and the limitations of our use of maximal isometric testing must be acknowledged. However, truly generalizable findings would have required quantifying the percentage of muscle activation during the loading phase of gait for each participant and then measuring muscle strength at this level. We chose our methods based on the reliability concerns raised above and to limit participant burden. Future researchers may wish to employ submaximal eccentric muscle testing to further assess the role of quadriceps in this patient population.

The secondary purpose of this study was to identify demographic, clinical, and biomechanical differences between individuals with knee OA who exhibited high impact loading and those who did not. We categorized participants based on the occurrences of HSTs as these have been implicated in high rates of loading and tibial accelerations [21,37] and have been suggested to be a major etiological factor in the development of OA [38]. Although we did lose a significant proportion of our original dataset (64%), we chose to make this rather discrete delineation to permit the examination of truly divergent groups that would be more difficult if using a continuous variable such as the KAM or magnitudes obtained from the Fz curve, or by categorizing individuals based on the number of HSTs they exhibited. Recognizing that many participants exhibited significant within-subject variability in terms of HST occurrence, we chose to take the conservative approach of dichotomizing our sample as it was the clearest way of separating the groups and testing our hypothesis. Since most of our participants exhibited HSTs at least some of the time, placing our strict criteria (either a significance of HST trials or absolutely no HST trials) increased our confidence in our group allocation. Though we do not believe that this limits our interpretation of the data, future researchers may wish to categorize participants based on the number of trials with HSTs, rather than the presence or absence of them.

Lower limb alignment was the only measured variable that was different between those who did and did not exhibit HSTs. Previous authors have suggested that malalignment mediates the effect of muscle strength on joint loading such that changes in the direction of force production with malalignment may create damaging forces within the knee [39,40]. Although strength was not found to be significant, results from the present study appear to support the importance of lower limb alignment on joint loading. Those with varus malalignment may experience increased impact loading, as evidenced by HST occurrence and higher maximum loading rates, due to altered structural mechanics. Though results from our study suggest a relationship between malalignment and altered loading patterns, our cross-sectional design limits the ability to identify cause-and-effect between these two variables. Clearly, further research is needed to examine further these disease-related phenomena.

Surprisingly, no differences were observed between groups with respect to freely chosen walking speed, vertical ankle velocity at initial contact, or knee flexion angles at initial contact or during loading — variables all suggested to be associated with impact loading and the presence of HSTs in healthy individuals [11,17,18,20,41,42]. We are

### Table 3

Demographic, clinical, and gait data in participants exhibiting heelstrike transients in most walking trials (n = 39) and those exhibiting no heelstrike transients (n = 35). Values are mean (sd) and the difference between the non-HST group and the HST group.

<table>
<thead>
<tr>
<th></th>
<th>Non-HST (n = 35)</th>
<th>HST (n = 39)</th>
<th>Diff (95% CI)</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (years)</td>
<td>64.7 (7.7)</td>
<td>64.0 (8.4)</td>
<td>0.7 (−3.0,4.4)</td>
<td>0.80</td>
</tr>
<tr>
<td>Gender (n)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Male</td>
<td>15</td>
<td>17</td>
<td></td>
<td>0.95</td>
</tr>
<tr>
<td>Female</td>
<td>20</td>
<td>22</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Height (m)</td>
<td>1.62 (0.08)</td>
<td>1.66 (0.09)</td>
<td>−0.04 (−0.07,0.01)</td>
<td>0.39</td>
</tr>
<tr>
<td>Mass (kg)</td>
<td>78.6 (14.5)</td>
<td>78.1 (15.4)</td>
<td>0.5 (−6.4,7.5)</td>
<td>0.72</td>
</tr>
<tr>
<td>KL grade (n)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>18</td>
<td>11</td>
<td></td>
<td>0.04</td>
</tr>
<tr>
<td>3</td>
<td>12</td>
<td>13</td>
<td></td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>5</td>
<td>15</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lower limb alignment (° varus)</td>
<td>2.5 (2.9)</td>
<td>4.6 (3.7)</td>
<td>−2.1 (−3.6, −0.5)</td>
<td>0.008a</td>
</tr>
<tr>
<td>WOMAC</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pain</td>
<td>8.4 (3.1)</td>
<td>7.4 (3.3)</td>
<td>1.0 (−0.5,2.4)</td>
<td>0.18</td>
</tr>
<tr>
<td>Function</td>
<td>27.6 (11.0)</td>
<td>25.1 (11.2)</td>
<td>2.6 (−2.7,7.8)</td>
<td>0.33</td>
</tr>
<tr>
<td>Quadriceps torque (Nm/kg)</td>
<td>1.28 (0.42)</td>
<td>1.39 (0.57)</td>
<td>−0.11 (−0.35,0.14)</td>
<td>0.38</td>
</tr>
<tr>
<td>Free pace walking</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Walking speed (m/s)</td>
<td>1.24 (0.16)</td>
<td>1.25 (0.18)</td>
<td>−0.01 (−0.09,0.06)</td>
<td>0.71</td>
</tr>
<tr>
<td>Vertical ankle velocity at initial contact (m/s)</td>
<td>−2.23 (0.81)</td>
<td>−2.48 (0.88)</td>
<td>−0.25 (−0.64,0.14)</td>
<td>0.20</td>
</tr>
<tr>
<td>Knee flexion at initial contact (°)</td>
<td>−4.64 (6.34)</td>
<td>−6.24 (6.92)</td>
<td>−1.60 (−4.68,1.40)</td>
<td>0.31</td>
</tr>
<tr>
<td>Maximum knee flexion during loading (°)</td>
<td>16.75 (8.19)</td>
<td>17.66 (7.22)</td>
<td>−0.93 (−4.61,2.76)</td>
<td>0.62</td>
</tr>
<tr>
<td>Fz peak magnitude (BW)</td>
<td>1.14 (0.08)</td>
<td>1.11 (0.12)</td>
<td>0.04 (−0.01,0.08)</td>
<td>0.15</td>
</tr>
<tr>
<td>Average loading rate (BW/s)</td>
<td>7.90 (2.77)</td>
<td>6.52 (2.02)</td>
<td>1.34 (0.26,2.49)</td>
<td>0.02</td>
</tr>
<tr>
<td>Time at Fz peak magnitude (% stance)</td>
<td>22.9 (4.3)</td>
<td>26.4 (4.6)</td>
<td>−3.5 (−5.5,1.4)</td>
<td>0.39</td>
</tr>
<tr>
<td>Max loading rate (BW/s)</td>
<td>10.94 (4.58)</td>
<td>24.87 (7.76)</td>
<td>−4.92 (−7.92,−1.93)</td>
<td>0.002a</td>
</tr>
<tr>
<td>Time at max loading rate (% stance)</td>
<td>6.25 (3.36)</td>
<td>4.86 (1.96)</td>
<td>1.39 (0.02,2.65)</td>
<td>0.03</td>
</tr>
</tbody>
</table>

*Denotes significance at p < 0.01.
not aware of any other studies that have examined correlates of HST occurrence in this patient population.

4.1. Limitations

The findings of this study should be viewed in light of some potential limitations. Firstly, all gait analyses were conducted in participants’ own low-heeled sneakers. This was chosen based on the ability to generalize outside the laboratory setting and to standardize as much as possible. It is known that the magnitude of the HST is dependent upon the compliance of the material contacting the ground [17]. Therefore, although it is unknown whether shoe type influenced the occurrence of HSTs in the present study, they may have decreased the overall rates of loading. Secondly, as indicated above, we chose to quantify “load” based on rates of loading and the presence/absence of HSTs, rather than other discrete measures such as the peak KAM or those obtained through musculoskeletal modelling. Although the KAM undoubtedly provides valuable information regarding disease pathomechanics, we believe that the measures used in this study provide additional insight into the role of loading in the disease process. Further, using only measures obtained from a force platform provides disease-specific outcome measures for researchers without access to costly three-dimensional motion capture systems. Finally, although a significant association was observed between HST occurrence and radiographic disease severity, a definitive association between the two constructs was not possible due to the cross-sectional nature. As a result, further research investigating the longitudinal effects of impact loading on articular cartilage degeneration is required.

5. Conclusion

Results from the current study indicate no significant relationship between the overall magnitude of quadriceps strength and rate of loading during walking, and adds to recent studies examining the effects of isolated quadriceps strengthening programs on discrete measures of loading such as the knee adduction moment [32–34]. Based on these findings, there has been a trend towards advocating strengthening of other muscle groups — for example, muscles of the hip — or protocols targeting functional movement patterns and/or neuromuscular retraining. Our findings of no significant relationship between quadriceps strength and rate of loading or HST presence further question the utility of isolated quadriceps strengthening in the treatment of knee OA and provide justification to examine the role of other lower limb muscles in knee joint loading and OA pathogenesis. Continuation of research into the role of muscle function on various knee joint loading parameters in those with knee OA will aid in the development of targeted rehabilitation strategies aimed at reducing disease development and progression.

6. Conflict of interest

None of the authors has any financial or personal relationships that would be deemed a conflict of interest.

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