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Tibiofemoral Load Magnitude and Distribution During Load Carriage

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Chronic exposure to high tibiofemoral joint (TFJ) contact forces can be detrimental to knee joint health. Load carriage increases TFJ contact forces, but it is unclear whether medial and lateral tibiofemoral compartments respond similarly to incremental load carriage. The purpose of our study was to compare TFJ contact forces when walking with 15% and 30% added body weight. Young healthy adults ($n = 24$) walked for 5 minutes with no load, 15% load, and 30% load on an instrumented treadmill. Total, medial, and lateral TFJ contact peak forces and impulses were calculated via an inverse dynamics informed musculoskeletal model. Results of 1-way repeated measures analyses of variance ($\alpha = .05$) demonstrated total, medial, and lateral TFJ first peak contact forces and impulses increased significantly with increasing load. Orthogonal polynomial trends demonstrated that the 30% loading condition led to a curvilinear increase in total and lateral TFJ impulses, whereas medial first peak TFJ contact forces and impulses responded linearly to increasing load. The total and lateral compartment impulse increased disproportionally with load carriage, while the medial did not. The medial and lateral compartments responded differently to increasing load during walking, warranting further investigation because it may relate to risk of osteoarthritis.

***Keywords*:** gait, vest-borne loads, knee joint contact forces

Osteoarthritis (OA) can cause pain and decrease quality of life; there are 32.5 million US adults with OA,[1] and this number is expected to rise to 78.4 million by 2040.[2] The average annual cost of OA in the United States is approximately 486 billion dollars, with up to 181 million work days lost.[1] Joint loading during daily activities, like walking and climbing stairs, circulates nutrients and promotes repair,[3–5] but chronic knee joint overloading is a risk factor for the onset and progression of OA.[3,5,6] Chronic exposure to elevated mechanical compressive forces may activate inflammatory pathways, production of cytokines, and activation of aggrecanases that catalyze destruction of cartilage extracellular matrix components and chondrocyte death.[3,5] In humans, lower extremity joints experience increased contact forces when there is excessive weight,[7–11] such as when soldiers carry heavy loads.[9–12] Therefore, understanding knee joint contact force magnitudes and distributions during load carriage could provide insight into how the knee joint articular environment is altered by this activity.

The tibiofemoral joint (TFJ) has medial and lateral compartments with different bone morphology and shape of the menisci,[13] coupled with differences in cartilage thickness and volume.[14] The medial compartment typically experiences larger peak forces and impulses during gait tasks than the lateral compartment[15,16] and has a higher rate of OA than the lateral compartment.[17,18] Previous knee joint models lump the contact forces between the compartments, providing total knee joint contact force estimates.[19–22] However, the distribution of contact forces across the medial and lateral compartments during cyclical loading tasks provides more precise estimates of tissue loads relevant to mechanisms of articular cartilage structural fatigue. Biomechanical models with compartment-specific TFJ contact forces are therefore needed to estimate these articular tissue loads due to added external mass. With these models, both first peak force and force impulse can be examined as each may provide unique information regarding OA occurrence and risk.[23,24]

Lower extremity gait mechanics change with load carriage,[9,12,25–31] with most of the adaptations occurring at the knee.[12,27,28,30] Knee flexion excursion increases during early stance with heavy load carriage.[12,27,28,30] The increase in knee flexion has been attributed to the lower extremity increasing shock absorption to mitigate the impact.[28] Concomitantly, greater knee flexion during load carriage lengthens the ground reaction force external moment arm to the knee[32] and increases the knee extension moment produced by the quadriceps,[12,32] and thus, increases TFJ contact forces.[31]

Peak total TFJ (tTFJ) contact forces during gait are known to vary as a function of body weight and added load,[15,33,34] but as body weight changes, TFJ contact forces often change by a different amount. Two weight loss studies report 4-[20] and 2-fold[35] decreases in knee joint contact forces compared to the lost weight in obese adults, respectively. Similarly, previous studies report a 2:1 ratio increase in first peak medial TFJ (mTFJ) contact forces to added weight.[31] But when examining relative values, load carriage of 15 and 30 kg resulted in a 18% and 36% increase in body weight, but only a 10.1% and 19.9% increase in first peak medial joint contact forces.[26] The relative increase in external load was not equivalent to the increase in relative mTFJ contact forces.[26] The previous load carriage studies all applied absolute external loads,[26,31] while 1 standard deviation of their participants' body mass in 2 studies were 13.4 kg[31] and 12.1 kg.[26] Applying an absolute mass to every participant is not experienced equally, such as a 15-kg load applied to a 65 kg person versus a 90-kg person, in which it is 23.1% and 16.7% relative mass added, respectively. Absolute mass would only be experienced equally if each participant was the same, which makes it difficult to determine the effect of added mass on contact forces. Additional studies of added mass relative to body weight on the magnitude and distribution of TFJ contact forces are necessary.

The purpose of this study was to compare TFJ contact forces and walking patterns when walking without load and while loaded

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with a weighted vest at 15% and 30% body weight, and to explore gait characteristics that may explain the differences in contact forces, if present. We hypothesized vest-borne loads relative to body weight will increase total, medial, and lateral TFJ contact forces and impulses in healthy young adults during gait, but there will be smaller increases in TFJ contact forces with progressively greater vest-borne loads relative to body weight due to gait changes used to decrease contact forces. Changes in joint contact forces can be unique to changes in added weight because joint contact forces are greatly influenced by muscle forces in addition to total body weight.[36,37]

Methods

Participants

The study was approved by the East Carolina University Institutional Review Board. An a priori power analysis using G*Power (version 3.1.9.6)[38] with a moderate effect size (η_p^2 = .08), α = .05, and β = 0.2 determined that 21 participants were necessary to detect differences between load conditions for this repeated measures design. The expected partial eta squared effect size was set at .08 based on previous literature determining large effect sizes for increase in peak mTFJ contact forces and mTFJ impulse when carrying a 20-kg load and moderate to large effect sizes for increase in peak mTFJ and lateral TFJ (lTFJ) contact forces when carrying 15- and 30-kg loads.[26,31] However, a sample of 24 healthy young adults (Table 1) were recruited to provide subsequent power for other hypotheses and to be conservative on previously unmeasured variables such as lTFJ contact force impulse. Participants were included if they were between 18 and 30 years old with a body mass index between 18 and 25 kg·m^{-2}, with no history of major lower extremity injuries or surgery and a leg length discrepancy of <2 cm. Every participant provided verbal and written consent prior to collection.

Procedure

A Health Survey and the Physical Activity and Readiness Questionnaire were administered over the phone prior to the participant's visit to the lab to determine whether the participant met the criteria for the study and whether the participant was able to perform the activity without consulting a physician. Written informed consent was obtained upon the participant's arrival to the Human Movement Analysis Lab (Department of Physical Therapy, East Carolina University). Height and weight were measured with a mechanical beam physician scale. Leg length was measured from greater trochanter to lateral malleolus with a tape measure. An unloaded Mir Pro or Mir Women's adjustable weighted vest (Mir) was fitted to the participant. Sixty-six reflective markers were applied to bony landmarks (acromioclavicular joints, iliac crests, anterior superior iliac spines, greater trochanters, femoral condyles, tibial plateaus, malleoli, first and fifth metatarsals, and distal tip of shoe) and segments with rigid clusters (posterior superior iliac spines and S2 cluster, thighs, and shanks) of the trunk, pelvis, and lower extremities to establish segment coordinate systems, define joint centers, and obtain motion data.[39] Markers on the trunk were placed on the vest rather than the skin or tight-fitting clothing. Three-dimensional marker data were collected at 200 Hz (Qualisys), and force data were sampled at 2000 Hz on an instrumented split-belt treadmill (Bertec).

Following warm-up and acclimation to the treadmill, each participant walked at 1.4 m·s^{-1} for 3 conditions: vest with no additional load, vest loaded to 15% body weight, and vest loaded to 30% body weight (Table 2) in that order. We did not randomize the conditions out of respect for the participant's time. Loading the vest in a sequential manner saved time during collection, because unloading the vest while the participant was wearing it was found to be especially difficult when piloting. We were also collecting expired gases to determine metabolic energy expenditure, and removing the metabolic equipment to unload the vest was burdensome and further increased collection time. We chose a standard speed to examine the effects of load without the confounding effects of variable speeds between participants and conditions. The vest was loaded with 1.36-kg steel bricks that were individually placed in the vest's pouches. Each condition was 5 minutes long, followed by a 5-minute rest break. Motion and force data were recorded in three 10-second intervals via Qualisys Track Manager during the last 2 minutes of each condition. The participants took a 5-minute break after each condition.

Processing

The final 10-second motion data file was trimmed in Qualisys Track Manager to each complete right stance phase so that we had 5 stance phases from each participant for each loading condition. The second to last 10-second motion data file was used to complete the 5 stance phases if they could not be extracted from the final 10-second motion file due to steps on both belts. A custom LabVIEW program[40] (National Instruments) used inputs of 5 stance phases for the right leg and then filtered marker and ground reaction force data with a fourth-order, zero-lag, low-pass, Butterworth filter and 15-Hz cutoff frequency.[40,41] These filtered data were used to calculate lower extremity moments and joint reaction forces via inverse dynamics (Visual 3D, C-Motion). Lower extremity sagittal plane kinematic data, computed joint moments, and joint reaction forces were input to a previously described musculoskeletal model

Table 1 Participant Demographics, Mean (SD)

	Females (n = 12)	Males (n = 12)	Total (N = 24)
Age, y	22.7 (1.6)	22.7 (3.0)	22.7 (2.3)
Height, m	1.70 (0.06)	1.80 (0.10)	1.75 (0.09)
Mass, kg	60.0 (6.6)	73.0 (7.0)	66.5 (9.4)
BMI, kg·m^{-2}	20.6 (1.5)	22.4 (1.3)	21.5 (1.6)

Abbreviation: BMI, body mass index.

Table 2 Load Added, Mean (SD)

	Females (n = 12)	Males (n = 12)	Total (N = 24)
15% load, kg	9.3 (1.4)	10.9 (1.1)	10.1 (1.5)
30% load, kg	18.1 (2.0)	22.1 (2.0)	20.1 (2.8)
Baseline impulse, N·s	**370 (42)**	**474 (47)**	**422 (69)**
15% impulse, N·s	**427 (48)**	**556 (64)**	**491 (86)**
30% impulse, N·s	**489 (54)**	**637 (73)**	**563 (98)**

Note: Baseline impulse was determined by taking the participant's body weight in newtons while standing statically and multiplying it by their mean stance time of that participant's no-load condition. 15% and 30% impulses were determined in the same way as the baseline but incorporated the added mass from the external load and the mean stance time for that condition.

of TFJ contact forces.[19,40,42] Estimates of gluteus maximus, hamstring, quadriceps, soleus, and gastrocnemius forces were derived using muscle moment arms as a function of lower extremity kinematics.[19,40,42–46] tTFJ compressive force was determined as the summed vertical and anterior–posterior components of the muscle forces and TFJ reaction forces acting perpendicular to the tibial plateau.[35,36] Total TFJ compressive force was then parsed to medial and lateral tibial compartments by applying the tTFJ compressive force to contact points on the medial and lateral tibial plateaus at 25% and 75% of subject-specific knee joint width measurements, respectively, in a manner that reproduced the net frontal plane knee joint moment.[40,42,47,48] Peak tTFJ contact force and impulse estimates using this model are within 3% and 7%, respectively, of in vivo gait data from an individual with an instrumented knee prosthesis, while peak mTFJ contact force and impulse are within 7% and 4% of in vivo gait data.[16,49] The primary dependent variables were first peak tTFJ, mTFJ, and lTFJ contact forces and impulses. Impulses were calculated over the time for each individual stance phase. We explored the role of muscle forces from the quadriceps, gastrocnemius, and hamstrings because muscle forces are a significant contributor to joint contact forces during walking.[36] Baseline impulse was determined by taking the participant's weight in newtons while standing statically and multiplying it by their mean stance time of that participant's no-load condition. The impulse for the 15% and 30% conditions was determined in the same way as the baseline but incorporated the added mass from the external load and the mean stance time for those conditions. We also explored peak knee flexion in early stance due to its relationship with the quadriceps forces but also because the kinematics may be associated with the primary dependent variables.

Analysis

A 1-way repeated measures analysis of variance ($\alpha = .05$) was used to compare the dependent variables in the right limb across the 3 (0%, 15%, and 30%) external loading conditions (SPSS, version 28). To confirm whether the within-subject factors had equal variance, Mauchly test of sphericity was used ($\alpha = .05$). If sphericity was violated ($P < .05$), Greenhouse–Geisser corrections were made to the degrees of freedom of the dependent variables to enable interpretation of the repeated measures analysis of variance.[50] Partial eta squared (η_p^2) within-subject effect sizes[51,52] were determined for each analysis of variance. Effect sizes were considered small (0.02), medium (0.08), or large (0.14).[52,53] Bonferroni corrected pairwise comparisons were used for post hoc analysis ($\alpha = .05$). Polynomial orthogonal trends were used to determine the presence of linear and quadratic trends ($\alpha = .05$), along with partial eta squared effect sizes of the trends, with a significant quadratic trend being used as an indicator for a curvilinear change between conditions.

Results

First peak tTFJ, mTFJ, and lTFJ contact forces increased directly with increased load carriage ($P < .001$; Table 3; Figures 1 and 2). From 0% load to 15% and from 0% to 30%, first peak tTFJ increased on average 316 N (16.0%) and 706 N (35.7%), resulting in 3.2:1 and 3.6:1 ratio increases in first peak tTFJ contact forces to weight added (Table 3 and Figure 2). First peak mTFJ increased 193 N (13.2%) and 405 N (27.6%), resulting in a 2.0:1 and 2.1:1 ratio increase, and first peak lTFJ contact forces increased 108 N (14.3%) and 263 N (35.1%), resulting in a 1.1:1 and 1.3:1 ratio increase, respectively (Table 3 and Figure 2). The statistical tests for linear trends in the peak TFJ contact forces were significant with large effect sizes, and the statistical tests for quadratic trends in the peak TFJ forcers were not significant (Table 3).

The tTFJ, mTFJ, and lTFJ impulses also increased directly with increased load carriage ($P < .001$; Table 3 and Figure 1). From 0% load to 15% and from 0% to 30%, tTFJ impulse increased on average 123 N·s (14.5%) and 278 N·s (32.7%), resulting in a 1.8:1 and 2:1 ratio increases in tTFJ impulse compared to standing impulse (Table 3). When parsed to the 2 compartments, mTFJ impulse increased 83 N·s (14.6%) and 169 N·s (29.8%), resulting in a 1.2:1 and 1.2:1 ratio increase; lTFJ impulse increased 41 N·s (14.2%) and 110 N·s (38.2%), resulting in a 0.6:1 and 0.8:1 ratio increase (Table 3). The statistical tests for linear trends with all TFJ impulses and quadratic trends with tTFJ and lTFJ impulses were significant with large effect sizes (Table 3).

Peak hamstring, first peak quadriceps, and peak gastrocnemius force increased with load carriage (Supplementary Table S1 and Supplementary Figure S1 [available online]). From 0% load to 15% and from 0% to 30%, peak hamstring force increased 34 N (7.6%) and 104 N (23.2%), peak quadriceps force increased 164 N (16.5%) and 407 N (40.8%), and peak gastrocnemius force increased 119 N (14.0%) and 253 N (29.9%), respectively (Supplementary Table S1

Table 3 TFJ Contact Forces and Impulses, Mean (SD)

Variable	No load	15% load	30% load	Main effects P (η_p^2)	Trend linear P (η_p^2)	Trend quadratic P (η_p^2)
First peak tTFJ, N	1978 (501)	2294 (550)**	2685 (692)**,‡	**<.001 (.853)**	**<.001 (.890)**	.189 (.074)
First peak mTFJ, N	1462 (329)	1655 (374)**	1867 (430)**,‡	**<.001 (.842)**	**<.001 (.884)**	.602 (.012)
First peak lTFJ, N	725 (242)	833 (271)**	988 (330)**,‡	**<.001 (.779)**	**<.001 (.834)**	.108 (.108)
TFJ impulse, N·s	851 (184)	974 (221)**	1129 (278)**,‡	**<.001 (.863)**	**<.001 (.874)**	**.014 (.233)**
mTFJ impulse, N·s	563 (132)	645 (162)**	731 (187)**,‡	**<.001 (.833)**	**<.001 (.849)**	.705 (.006)
lTFJ impulse, N·s	288 (108)	329 (122)**	398 (154)**,‡	**<.001 (.789)**	**<.001 (.825)**	**.009 (.262)**

Abbreviations: ANOVA, analysis of variance; lTFJ, lateral TFJ; mTFJ, medial TFJ; η_p^2, partial eta squared; TFJ, tibiofemoral joint; tTFJ, total TFJ. Note: Means (SD) for first peak and impulses of the TFJ. Repeated-measures ANOVA results for main effects, Bonferroni pairwise comparisons, and post hoc quadratic trend analysis. No load is walking with an unloaded weighted vest, 15% load is walking with the weighted vest loaded at 15% body weight, and 30% load is walking with the weighted vest loaded at 30% body weight. Bolded text highlights significant P values and large effect sizes.
**$P \le .005$ compared with no load. ‡$P \le .005$ compared with 15% load.

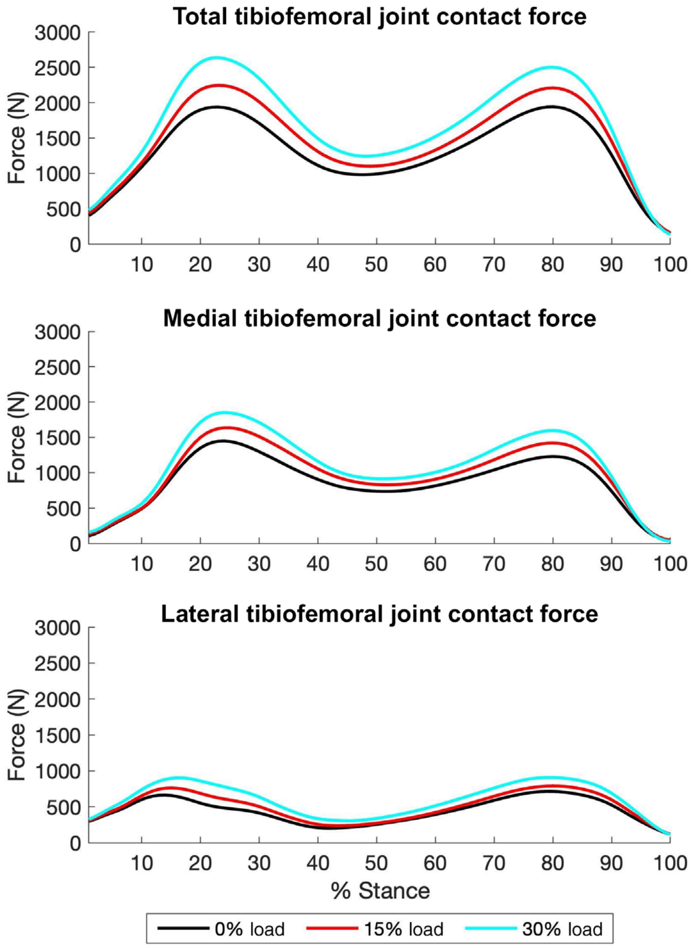

Figure 1 — Total, medial, and lateral tibiofemoral joint contact forces during stance phase of gait under 3 loading conditions. The first peak of total, medial, and lateral tibiofemoral joint contact forces and impulse increased across conditions ($P < .001$). Impulse was calculated over nonnormalized time. Lines represent the means across all participants.

[available online]). Hamstrings, quadriceps, and gastrocnemius impulse increased with increased load carriage (Supplementary Table S1 and Supplementary Figure S1 [available online]). From 0% load to 15% load and from 0% to 30%, hamstring impulse increased 10 N·s (12.1%) and 26 N·s (32.5%), quadriceps impulse increased 42 N·s (16.2%) and 96 N·s (38.1%), and gastrocnemius impulse increased 29 N·s (13.7%) and 67 N·s (31.8%; Supplementary Table S1 [available online]). Peak knee flexion angle during early stance increased directly with load (Supplementary Table S2 [available online]). From 0% to 15% load and from 0% to 30%, peak knee flexion increased 0.8° and 2.0° on average (Supplementary Table S2 [available online]).

Discussion

The purpose of this study was to compare TFJ contact forces and walking patterns when walking without load and while loaded with a weighted vest at 15% and 30% body weight. tTFJ, mTFJ, and lTFJ contact forces and impulses significantly increased with increasing load. Relative to body weight, all increases in peak TFJ contact forces were disproportionately greater than weight added. In contrast to our hypothesis, total impulse disproportionately increased at the 30% condition due to increased quadriceps and gastrocnemius force contributions. mTFJ impulses responded in a linear fashion to increasing load carriage while the lateral impulses curvilinearly increased with the 30% loading condition.

This study is not without limitations, such as the conditions for this study were not randomized or counterbalanced due to time constraints from loading and unloading the vest; therefore, fatigue and learning effects may be present. Previous studies have counterbalanced loading conditions,[12,26,54] and load carriage does induce metabolic fatigue,[55] but other studies did not find interactions of load carriage and fatigue on lower extremity mechanics.[56–59] Also, previous studies tend to examine load carried over long distances or for long periods of time, while we used intentionally short wear times for young and healthy participants to limit exertion effects with 5-minute breaks between load conditions. Though our experimental design does not explicitly control for the possibility of carryover effects in our results, we believe load manipulation is a more likely explanation for the differences observed between conditions in this study. Additionally, the model's lateral force estimates have not been formerly validated against in vivo contact forces. However, peak lateral TFJ contact forces in our baseline condition (ie, 0% load) were consistent with reported in vivo lateral TFJ contact force peaks of 556 to 871 N from an instrumented prosthesis while walking without a load.[16,60] There are no contact forces from an instrumented prosthesis to compare for 15% and 30% loadings in the literature. Finally, trunk markers were placed directly on the vest, and the length of the vests caused some difficulty in pelvis marker visualization during the 30% condition.

Increasing or decreasing mass can affect TFJ contact forces during walking. TFJ contact forces during walking with added load in this study were similar to those reported in the literature.[16,26,31] In addition to absolute force changes, we used an absolute ratio perspective (ie, absolute ratios of increased TFJ contact forces to weight added) in our analysis. The 3.2:1 and 3.6:1 ratio increases in absolute first peak tTFJ contact forces to weight added were greater than the 2:1 ratio of reduced TFJ contact forces relative to weight loss in obese adults reported by DeVita et al[35] but less than the 4:1 reduced TFJ contact forces reported by Messier et al[20] for obese adults with OA who lost weight. These conflicting results suggest that knee joint load responses to weight gain or mass added may not be the direct opposite to responses to weight loss. We also need to acknowledge that the differences between groups being compared (ie, healthy [present study], obesity, and OA) may account for differences in results. Individuals who are obese often have unique anthropometrics that can affect gait mechanics,[61,62] and people with OA have gait mechanics which can be affected by stiffness, symptoms, or swelling.[63]

In addition to peak TFJ contact forces increasing with external loads, stance time increased 5 and 14 ms with increasing load, which affected impulse. Unlike peak tTFJ contact force, total impulse was described by a significant quadratic trend, indicating that as the relative load increased linearly, there was a curvilinear increase in the impulse. This is contrary to our hypothesis and potentially due to the short-term exposure of the external loads and not having people with knee OA participate who may use a different technique to attenuate the external loads.[23] If impulses increase more rapidly at higher than lower loads, then chronic exposure to heavy loads in the military may increase the risk of developing OA or patellar tendinopathy.[31,64]

Medial and lateral tibiofemoral compartment first peak contact force increased with increasing load, similarly to previous

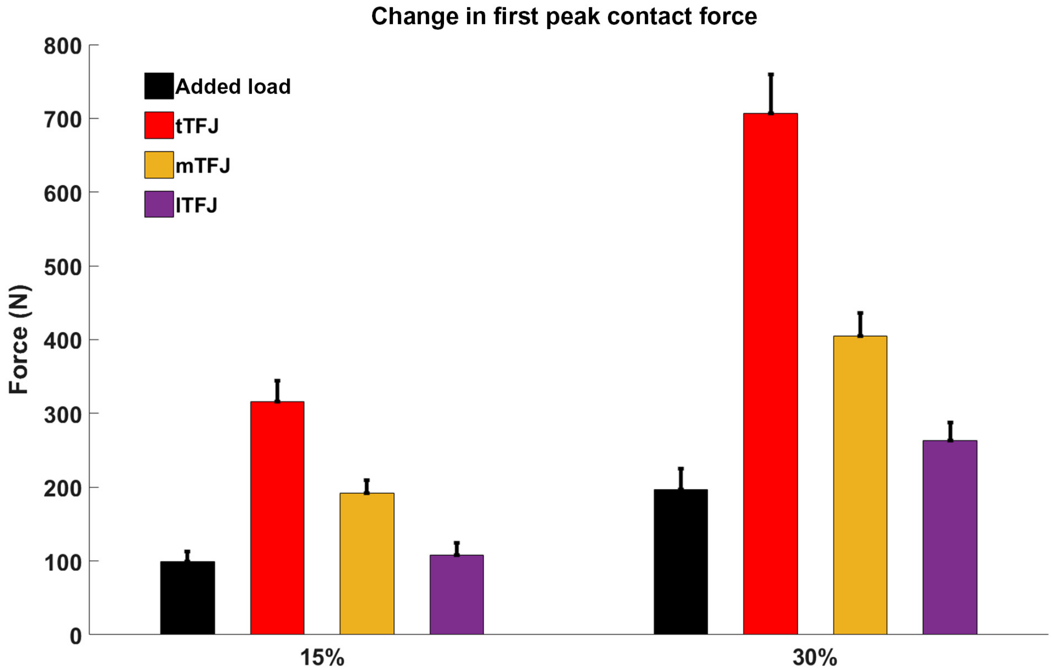

Figure 2 — Mean with standard error for change in first peak contact force for tTFJ, mTFJ, and lTFJ contact forces from 0% to 15% added load (left bars) and 0% to 30% added load (right bars), along with added weight for each condition (black bars). The contact forces are plotted on the same scale as the time-series data to show magnitude differences between the conditions and knee compartments. lTFJ indicates lateral TFJ; mTFJ, medial TFJ; TFJ, tibiofemoral joint; tTFJ, total TFJ.

studies.[26,31] We found a 2.0:1 and 2.1:1 ratio increase (15% and 30% condition, respectively) in first peak mTFJ contact force increase to weight added, which is similar to the 2:1 ratio previously reported[31] during load carriage of 20 kg (~26% body weight). Also similar to previous findings, mTFJ impulse increased with increasing load, with 1.2:1 ratios of mTFJ impulse to standing impulse for both conditions.[31] Contrary to our hypothesis, the peak mTFJ contact forces and impulses had a linear response from load carriage of 15% to 30% based on the consistency in absolute ratios and a significant linear trend with large effect size. Our results with first peak mTFJ contact forces and impulses do not suggest an attenuation of contact forces as external load increases. For the lateral compartment, we found that first peak lTFJ contact forces increased with increasing load. Despite the large increase from the 15% to 30% condition, peak lTFJ contact forces were best described by a linear trend. However, the lTFJ impulses demonstrated a significant quadratic trend. The differences in first peak contact forces and impulses between the medial and lateral compartments as the external loading increases require further investigation with greater loads to test whether each compartment has an independent response to load.

The quadriceps forces had a unique response to greater load, which were consistent in curve shape with the literature.[35,42] There was a 1.7:1 and a 2.1:1 increase in peak quadriceps force to weight added, with the 30% condition inducing a 40% increase in peak quadriceps force. The quadriceps exhibited 2 to 3 times greater peak force than the hamstrings, subsequently providing a greater contribution to first peak tTFJ contact force, which is consistent with the literature.[35,36,42] Therefore, peak quadriceps force was a contributing factor that helps explain the change in first peak tTFJ contact force ratio from 3.2:1 to 3.6:1 between the 15% and 30% load conditions.

The curvilinear increase with tTFJ impulse is most likely due to the ratios of quadriceps impulse and knee flexion increases across the conditions. The quadriceps impulse to standing impulse was 0.6:1 (15%) and 0.7:1 (30%). The width of the quadriceps peak at the 30% loading condition appears to be wider than the other 2 conditions in addition to having a larger magnitude (Supplementary Figure S1 [available online]), thus increasing the tTFJ impulse at the 30% condition. Also, there was a greater peak knee flexion during the 30% condition compared with the 15% condition, albeit only a 1.2° difference and likely not clinically significant and within the range of measurement error.[65] The greater peak knee flexion corresponded to the increase in peak quadriceps impulse and is consistent with the literature.[12,25,28,30,31] The increase in knee flexion also may have a role in energy absorption through a longer eccentric quadriceps action.[28] The increase in peak knee flexion occurring in the 30% condition (Supplementary Table S1 [available online]) likely increased the external ground reaction force lever arm from the knee joint center, which explains the change in the absolute ratio and why the relative increase in tTFJ contact force and quadriceps force was greater during the 30% condition. The differences in knee flexion between conditions were small but still may have been a contributing factor to the difference in knee joint contact forces.

The medial and lateral compartments responded differently to the loading conditions in this study, with the lateral compartment having a greater relative increase in impulse than the medial compartment during the heaviest loading condition. It is unclear whether this relationship between medial and lateral compartments would continue with greater load magnitudes or with prolonged load carriage, but there have been previous findings showing different responses in cartilage of the medial and lateral compartments after weight loss.[66] The curvilinear increase in tTFJ impulse may indicate that long periods of heavy load carriage may put the knee at risk. The cumulative overloading could initiate the maladaptive mechanical transduction pathway that leads to inflammation and the activation of cartilage degradation enzymes,[3,5] increasing risk of OA onset or progression. Similar to previous studies,[67,68] the present data suggest that impulse can be a strong indicator of changes in the knee joint environment during load carriage and that the 2 knee compartments respond differently to load.

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