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ORIGINAL RESEARCH

Electromyography-Informed Estimates of Joint Contact Forces Within the Lower Back and Knee Joints During a Diverse Set of Industry-Relevant Manual Lifting Tasks

Felicia R. Davenport,[1] Jennifer K. Leestma,[1,2] Adriana Staten,[3] Krishan Bhakta,[1] Joshua Fernandez,[1,2] Anirban Mazumdar,[1,2] Aaron J. Young,[1,2] and Gregory S. Sawicki[1,2,4]

[1]Woodruff School of Mechanical Engineering, Georgia Institute of Technology, Atlanta, GA, USA; [2]Institute for Robotics and Intelligent Machines, Georgia Institute of Technology, Atlanta, GA, USA; [3]Wallace H. Coulter Department of Biomedical Engineering, Georgia Institute of Technology, Atlanta, GA, USA; [4]School of Biological Sciences, Georgia Institute of Technology, Atlanta, GA, USA

Repetitive manual labor tasks involving twisting, bending, and lifting commonly lead to lower back and knee injuries in the workplace. To identify tasks with high injury risk, we recruited N = 9 participants to perform industry-relevant, 2-handed lifts with a 11-kg weight. These included symmetrical/asymmetrical, ascending/descending lifts that varied in start-to-end heights (knee-to-waist and waist-to-shoulder). We used a data-driven musculoskeletal model that combined force and motion data with a muscle activation-informed solver (OpenSim, CEINMS) to estimate 3-dimensional internal joint contact forces (JCFs) in the lower back (L5/S1) and knee. Symmetrical lifting resulted in larger peak JCFs than asymmetrical lifting in both the L5/S1 (+20.2% normal [$P < .01$], +20.3% shear [$P = .001$], +20.6% total [$P < .01$]) and the knee (+39.2% shear [$P = .001$]), and there were no differences in peak JCFs between ascending versus descending motions. Below-the-waist lifting generated significantly greater JCFs in the L5/S1 and knee than above-the-waist lifts ($P < .01$). We found a positive correlation between knee and L5/S1 peak total JCFs ($R^2 = .60$, $P < .01$) across the task space, suggesting motor coordination that favors sharing of load distribution across the trunk and legs during lifting.

Keywords: lumbar spine, injury risk, musculoskeletal modeling, internal joint loading

Workplace musculoskeletal disorders, such as low back pain and osteoarthritis in the knees can be physically and financially burdensome to members of the working class and will continue to have a negative impact on society as they age.[1,2] One of the leading causes of workplace injuries is overexertion in tasks, such as manual materials handling (MMH), manufacturing, and patient transport.[3] Workers in manual labor professions may perform repetitive, asymmetric, and physically taxing movements[4] such as twisting, pushing, or bending while handling heavy loads[5–7]—putting them at risk for acute injury when exposed to extreme peak loading or chronic overuse injury with prolonged exposure to joint and tissue loading.[8] Common approaches to characterize and mitigate injury risk in manual lifting have relied on metrics derived from measures of joint kinematics and kinetics. Factors such as varying the mass of a lifted object,[9–11] the size of a lifted object,[12] lifting speed,[10] lifting technique,[12,13] stance width and foot position,[12,14] lowering versus lifting,[15,16] and symmetry versus asymmetry (turning) all play significant roles in understanding their impact on musculoskeletal loading.[10,11,13,15,17,18] In sum, external loads on the limb-joints during MMH tasks have been well characterized and helped provide best practices to guide lifting techniques that can help avoid injuries.

Characterizing internal loads on underlying musculoskeletal structures may provide a more direct assessment of injury risk than external measures based on inverse dynamics. Joint contact forces (JCFs), potential biomarkers for osteoarthritis onset due to structural changes in cartilage in ACL injuries,[19–21] and correlated to low back injury risk[22] could reveal key underlying changes in cartilage and bone health. JCFs quantify the internal loading at bone-to-bone interfaces necessary to support the forces produced by surrounding muscles, ligaments, and tendons that are necessary to counter external loads.[23] Despite their potential utility, JCFs are impossible to measure noninvasively in humans, with only a few examples of direct *in vivo* measurements of JCFs.[24,25] Furthermore, mechanical analyses that attempt to map net external forces and moments to internal joint loads are not straightforward and require estimates of how antagonistic muscle forces are coordinated.[26] Fortunately, advances in computational biomechanics make it possible to utilize musculoskeletal modeling practices to reproduce MMH tasks in simulation[27,28] and estimate JCFs.[29–33] Electromyography (EMG)-informed solvers can reduce errors in JCF estimation from purely simulated muscle activation inputs by accounting for muscle co-activation,[34,35] which is a critical piece of the puzzle

Leestma https://orcid.org/0000-0002-4435-4121
Bhakta https://orcid.org/0000-0002-5708-9304
Fernandez https://orcid.org/0000-0002-9177-361X
Mazumdar https://orcid.org/0000-0002-3457-282X
Young https://orcid.org/0000-0002-5376-2258
Sawicki https://orcid.org/0000-0002-5588-6297
Davenport (fdavenport6@gatech.edu) is corresponding author, https://orcid.org/0000-0002-1186-0490

given that muscle action typically accounts for well over half the joint contact load.[36,37] These advanced computational tools make it possible to move "outside-in" and evaluate tissue-level mechanics in the context of highly dynamic movements—enabling monitoring of a new set of "local" biomarkers that could offer more precise injury prevention measures.

The purpose of this study was to extend limb-joint analyses and establish a framework to examine how external loads resolve as 3-dimensional JCFs *inside limb joints* during MMH tasks. To bridge this gap, we designed a comprehensive study protocol comprised of a diverse set of lifting tasks spanning a range of start-end heights and across a range of lift symmetries. In these conditions, we collected full-body motion data using high-speed motion capture, external ground reaction forces, and electromyography from a carefully selected set of muscles around the back and knee. We paired these experimental data with state-of-the-art computational modeling, simulation, and optimization techniques to evaluate lower back (L5/S1) and knee joint loads during lifts that varied starting and ending heights and lift symmetry, providing insight into potential "hot spots" for injury over a wide range of possible lifting situations. We hypothesized that asymmetrical,[38,39] ascending,[10] and below the waist lifts[11,40] would induce the highest JCFs at both the L5/S1 and knee (H1–H3). We expected JCFs at the knee and L5/S1 joints would be proportional to each other,[9,10,40–42] independent of the lifting task (H4)—reflecting a motor control strategy to avoid overloading the L5/S1 or knee by distributing loads across the limb joints. Studying the effects of MMH tasks on multijoint internal joint loading will enable a comprehensive mapping from task space to injury risk during industry-relevant lifting tasks. In doing so, we hope to provide a greater awareness of the susceptibility to eventual tissue and bone-damaging injuries in the workplace that can be used as a prophylactic tool to avoid prolonged exposure to critical loading in the workplace.

Methods

Participants

This study was approved by the Georgia Institute of Technology Institutional Review Board, and all participants provided signed consent prior to data collection. We enrolled 9 participants (7 males and 2 females; weight: 80.7 [15.0] kg; height: 178.7 [11.1] cm; age: 25.1 [2.9] y). Participants were excluded if they had a history of debilitating injuries from neurological, musculoskeletal, or cardiovascular conditions that prohibited them from successfully performing the manual labor tasks in this study.

Experimental Design

All participants performed 24 different lifts with varied lift start–end heights and degrees of twisting (Figure 1). Start–end height combinations included knee-to-waist (KW), waist-to-knee (WK), shoulder-to-waist (SW), and waist-to-shoulder (WS). Symmetric lifting had 0° of twisting, while asymmetric lifting could be with 90° or 180° of twisting, with turns centered around a neutral posture (ie, 180° lifts started and ended at ± 90° from neutral). Across conditions, participants lifted a 25 lb. (11.34 kg) dumbbell at their preferred lifting speed to fixed shelf positions at knee (17.8 cm [7 in]), waist (72.4 cm [28.5 in]), or shoulder (133.5 cm [52.5 in]) height.

Participants wore a full-body reflective marker set to record segment positions (Vicon, 100 Hz). We collected data from 16 surface EMG sensors (Delsys, Natick, Massachusetts, 2000 Hz). EMG sensors were placed on the right leg muscles (tibialis anterior, lateral gastrocnemius, rectus femoris, vastus medialis, vastus lateralis, bicep femoris, and semitendinosus), and bilaterally on the torso muscles (erector spinae, latissimus dorsi, external obliques, and right-only rectus abdominis; Figure 1B). Maximum voluntary contractions (MVCs) were performed by each participant

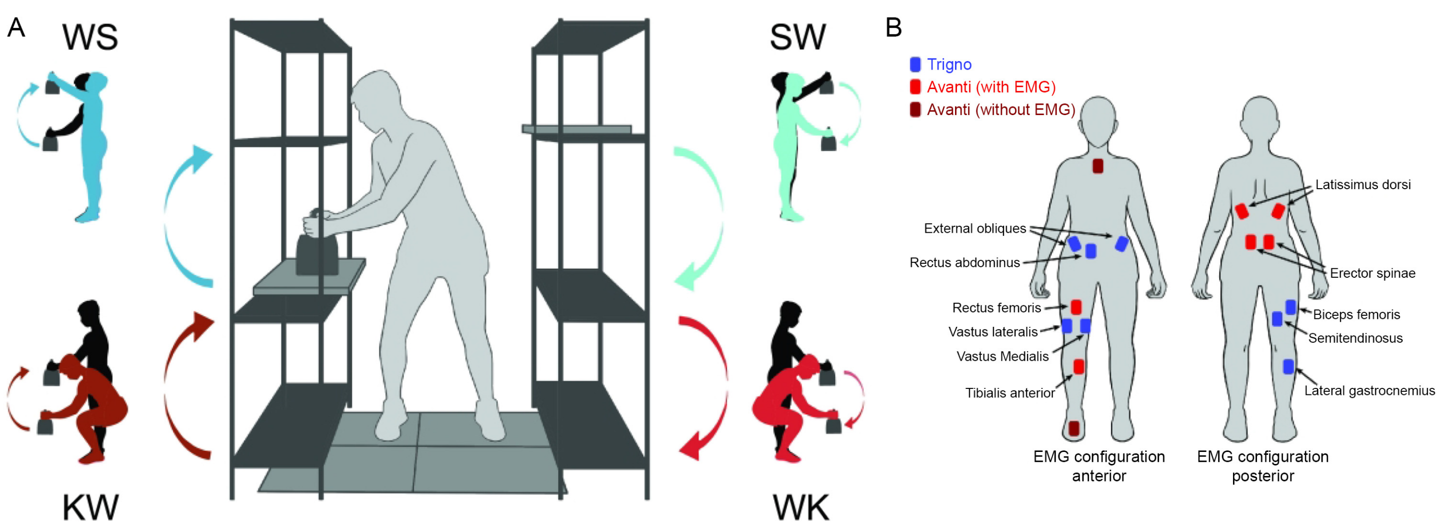

Figure 1 — (A) Manual lifting conditions visualized by starting shelf height and target shelf position of the weight. During 0° (symmetric) lifts, a single shelf was oriented directly in front of participants and weight was lifted to-and-from various heights. For both 90° and 180° (asymmetric) lifting, tasks were performed starting from both sides of the body: left-starting lifts (clockwise rotation) and right-starting lifts (counterclockwise rotation). In 90° lifts, shelves were offset 45° from neutral. In 180° lifts, 2 shelves were placed on opposite sides of the participants. The top, blue arrows represent above-the-waist lifting conditions: SW in light blue and WS in dark blue. The bottom, red arrows represent below-the-waist lifting conditions: KW in dark red and WK in light red. (B) Placement of surface EMG sensors. Muscles were selected to capture activity from muscle groups' salient to the targeted joints under study: lower back (L5/S1) and knee. Anterior and posterior muscle groups were collected for the torso and legs, but only the right leg was instrumented with sensors. EMG indicates electromyography; KW, knee-to-waist; SW, shoulder-to-waist; WK, waist-to-knee; WS, waist-to-shoulder. (Color figure online)

preceding the protocol to collect the maximum muscle activations for normalization.[43,44] We used SENIAM to inform sensor placement and MVC collection.[45] We collected ground reaction forces under each foot and each shelf (Bertec, 2000 Hz).

Data Analysis

Raw EMG signals were band-pass filtered (20–400 Hz, second-order Butterworth), full-wave rectified, low-pass filtered (6-Hz, fifth-order Butterworth), and half-wave rectified to create positively signed envelopes. Finally, the EMG signals were normalized using the peak MVC value across 3 MVC trials. We used the rate of change of ground reaction force from force plates on the shelves with a threshold value of 0.10 N/s to detect the start–end position of the dumbbell and define the initiation-termination time of each lift.

We implemented 2 OpenSim models (version 4.0 [SimTK][30]) to simulate experimental lifting trials: Full-Body Running model[46] for knee-joint analyses and Lifting Full Body model[29] for L5/S1-joint analyses. The Full-Body Running model contains 12 segments with 92 trunk and lower limb musculotendon actuators, while the Lifting Full Body model contains 30 segments with 238 torso musculotendon actuators. The Full-Body Running and Lifting Full Body models have a high resolution of knee and back muscles, respectively, providing representative muscle activations and forces from muscles which we collected EMG. To accurately simulate our unique lifting conditions, we modified the range-of-motion constraints of joints within the arms, torso, pelvis, and legs to allot freedom for the model skeleton to match the participants' movements. Additionally, to account for the external load contributions of the 25 lb. weight to internal musculoskeletal states, we added half the dumbbell mass to each hand using OpenSim *WeldJoint*.

Next, we fitted the 2 OpenSim models to participant-specific anthropometry taken from captured static pose data using the Scaling Tool. We used the Inverse Kinematics Tool to calculate joint angles. OpenSim's Inverse Dynamics Tool computed joint kinetics for each participant incorporating the effects of external loads. We used the Muscle Analysis Tool to solve for musculotendon unit lengths and muscle moment arms.

Then, we used the Calibrated EMG-Informed Neuromusculoskeletal Modeling Toolbox (CEINMS)[34] to perform optimization (ie, simulated annealing) and reduce error in joint and muscle-level property estimations. First, the CEINMS Hybrid mode optimized muscle activity over those muscles without experimental data inputs. Then, we used both the experimental and simulated muscle excitation signals, musculotendon lengths, and muscle moments to calibrate the model by reducing the error between estimated and experimental joint moments. The following objective function weighting parameters were inspired by those utilized in similar analyses by Molinaro et al[18] to balance muscle activations and torque-tracking performance: torque tracking (α) = 10,000, muscle excitation minimization (β) = 10, and EMG tracking (γ) = 1000.[18]

Finally, we used the OpenSim Joint Reaction Analysis Tool, taking CEINMS simulated output, to calculate the JCFs. From-time series data, we computed peak and integrated values using custom MATLAB scripts. We divided JCFs into normal, shear, and total components. We calculated shear and total JCFs using the Euclidean norm of the anterior-posterior and mediolateral shear forces and the Euclidean norm of the normal and shear forces, respectively. We normalized JCFs by the product of participant mass and acceleration of gravity to minimize the effects of performance variance across participants. To avoid duplication of data in asymmetrical lifting conditions (90° and 180° turns), we compared the forces in clockwise (left-starting) and counterclockwise (right-starting) directed lifts and analyzed/presented the trial with the maximum JCF (ie, the "worst case"). We reported right knee JCFs, from the leg instrumented with EMG sensors.

Statistical Analysis

We employed a 2-way repeated-measures analysis of variance to test for significance of lift symmetry, direction, and start-to-end height on each component of the JCFs (H1–3). Subsequently, we performed a *post hoc* Bonferroni pairwise multivariate comparison test to determine statistical significance of differences in JCFs across lifting conditions with the threshold for significance set at $\alpha = .05$ (Minitab–Penn State University). We used a linear regression model to compute correlation coefficients between the L5/S1 and knee JCFs (MATLAB; H4).

Results

Participants used different postures across lift conditions. Joint kinematics indicated asymmetric lifts had greater L5/S1 flexion (sagittal plane), lateral bending (frontal plane), and axial rotation (transverse plane) than symmetric lifts. Peak right knee flexion (sagittal plane) was less in asymmetric lifts than symmetric lifts across conditions (Supplementary Figure S1 [available online]). In addition, participants used different kinetic strategies[47] to perform lifting tasks (Figure 2). The peak (Supplementary Figure S2 [available online]) and integrated (Supplementary Figure S3 [available online]) flexion moments about the L5/S1 and right knee joints were smaller in asymmetric lifts (90° and 180°) compared with symmetric lifts (0°). The peak (Supplementary Figure S2 [available online]) and integrated (Supplementary Figure S3 [available online]) lateral bending and axial rotation moments about the L5/S1 joint were greater in asymmetric lifts than symmetric lifts.

Muscle activations are an important variable in determining muscle forces, the primary contributor to JCFs.[36,47] Across lifting conditions (Supplementary Figure S4 [available online]), the external obliques, vastus medialis, biceps femoris, and semitendinosus all had greater peak normalized muscle activations (EMG) in asymmetric (90° and 180°) versus symmetric lifts (0°). In contrast, the erector spinae, rectus femoris, and lateral gastrocnemius, all had greater peak normalized EMG in symmetric (0°) versus asymmetric lifts (90° and 180°). Latissimus dorsi and rectus abdominis muscles had negligible differences in their normalized EMG across lifts.

In the L5/S1 joint, asymmetric lifts (90°/180°) resulted in lower normal, shear, and total JCFs when compared with symmetric lifts (Figure 3A). Peak normal forces were 7.2% ($P = .046$) and 20.2% ($P < .01$) lower in asymmetric lifts at 90° and 180° versus symmetric lifts (0°), respectively. Peak shear forces were 20.3% ($P = .001$) and 14.3% ($P = .032$) lower for 90° and 180° lifts versus symmetric lifts (0°). Peak total forces were 7.3% ($P = .032$) and 20.6% ($P < .01$) lower for 90° and 180° lifts versus symmetric lifts (0°; Figure 4A, Table 1). Peak normal, shear, and total JCFs were not different in ascending (KW/WS) versus descending (WK/SW) lifts in the L5/S1 joint (Figure 5A). Peak normal, shear, and total JCFs were 21.0%, 49.3%, and 23.3% higher in below-the-waist (KW/WK) than above-the-waist (SW/WS) conditions ($P < .01$) in the L5/S1 joint (Figure 6A).

In the right knee, 180° asymmetric lifts resulted in lower normal, shear, and total JCFs when compared with 0° and 90° lifts (Figure 3B). Peak normal forces were 17.5% lower in 180° versus 90° lifts ($P < .01$), peak shear forces were 39.2% lower in 180° vs

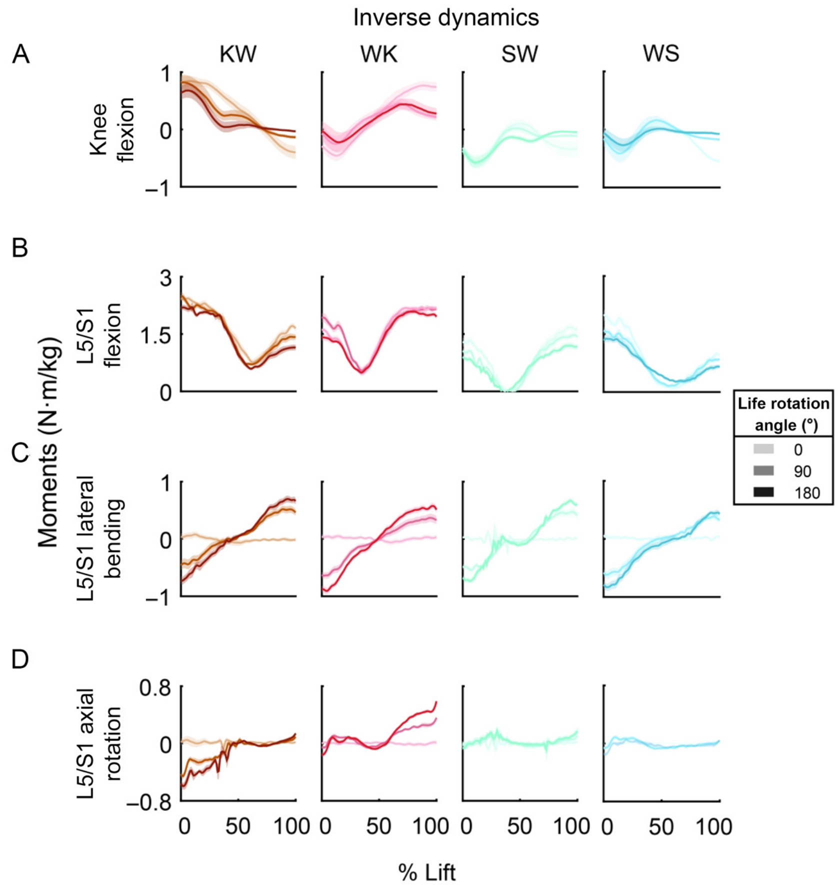

Figure 2 — The across-participant averaged time series of L5/S1 and knee kinetics of knee flexion (A), L5/S1 flexion (B), lateral bending (C), and axial rotation (D) during KW, WK, SW, and WS lifting conditions. The color opacity of each data line increases with degree turn of lift: from 0° (lightest) to 180° (darkest). The haze surrounding each average represents the standard error of the mean. A comparative analysis was performed across each asymmetric lift (90° and 180°) condition to select data from the starting side (right or left) that resulted in greater (worse) joint moments. Knee joint moments were reported from the right, instrumented leg. Positive joint moments for the L5/S1 and knee flexion components represented greater joint flexion. Axial rotation and lateral bending components of the L5/S1 joint were positive moving away from the starting side. KW indicates knee-to-waist; WK, waist-to-knee; SW, shoulder-to-waist; WS, waist-to-shoulder.

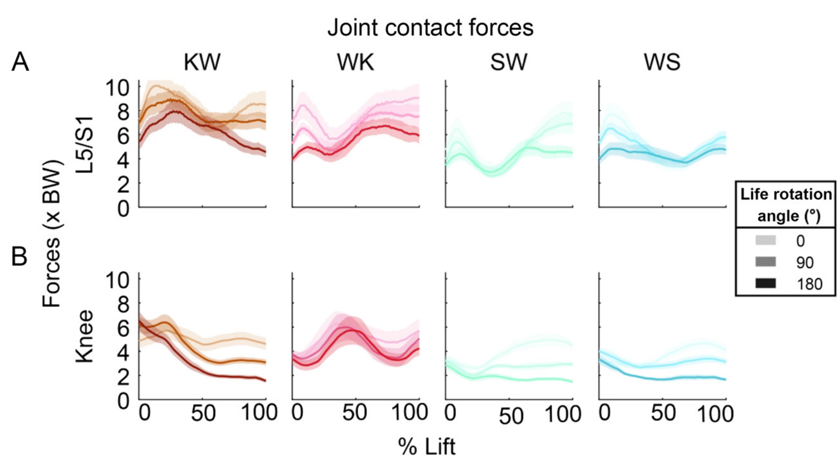

Figure 3 — The across-participant averaged time series for total JCFs in the L5/S1 joint (A) and knee (B). The 4 panels in each figure illustrate data of the lifting conditions: KW, WK, SW, and WS. The color opacity of each data line increases with degree turn of lift: from 0 (lightest) to 180° (darkest). The haze surrounding each average represents the standard error of the mean. JCFs indicates joint contact forces; KW, knee-to-waist; SW, shoulder-to-waist; WK, waist-to-knee; WS, waist-to-shoulder.

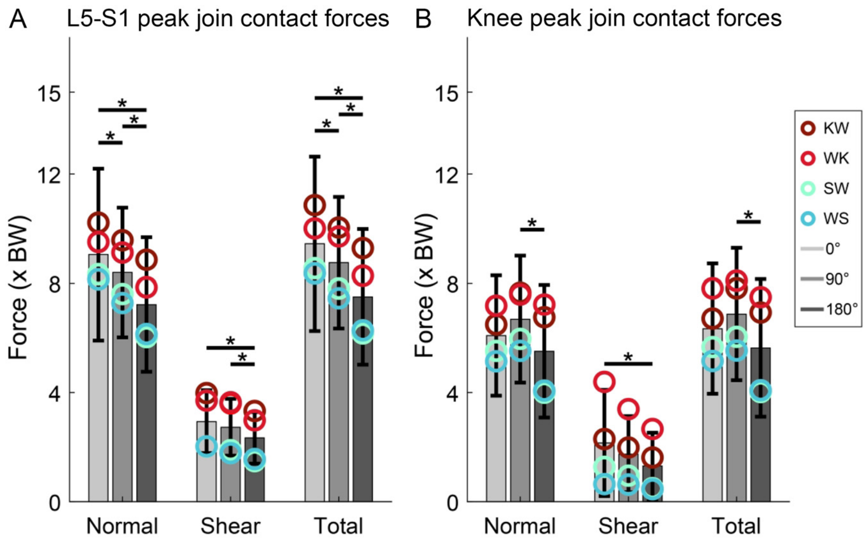

Figure 4 — The across-participant averaged peak normal, shear, and total JCFs in the knee and L5/S1 joint as explored in the first hypothesis (H1). The bars represent the JCF value averaged across the 4 lift conditions (KW, WK, SW, and WS) for each of the 3 lift degrees (0°, 90°, and 180°) in light to dark gray. Independent values for each lift condition are labeled as circles in the corresponding color (KW: red, WK: pink, SW: dark blue, and WS: light blue). Significantly different groups are denoted with an asterisk. Statistical significance across all conditions concluded when $\alpha = .05$. JCFs indicates joint contact forces; KW, knee-to-waist; SW, shoulder-to-waist; WK, waist-to-knee; WS, waist-to-shoulder. (Color figure online)

Table 1 Effect of Lifting Rotation and Start-to-End Height on Knee and Lower Back (L5/S1) Peak Total Joint Contact Forces (×BW)

	Start-to-end heights			
Rotation, °	**KW**	**WK**	**SW**	**WS**
Knee				
0°	6.71 (2.10)	7.81 (3.53)	5.67 (1.34)	5.16 (1.22)
90°	7.82 (2.28)	8.09 (3.16)	6.04 (1.70)	5.54 (1.44)
180°	6.93 (2.22)	7.49 (3.04)	4.02 (1.03)	4.08 (1.03)
Lower back (L5/S1)				
0°	10.86 (3.49)	10.01 (3.40)	8.55 (2.95)	8.36 (2.71)
90°	10.04 (2.53)	9.71 (2.62)	7.81 (2.00)	7.45 (1.52)
180°	9.29 (2.68)	8.28 (2.38)	6.16 (1.92)	6.28 (1.57)

Abbreviations: KW, knee-to-waist; SW, shoulder-to-waist; WK, waist-to-knee; WS, waist-to-shoulder; ×BW, ×body weight.
Note: Knee and lower back (L5/S1) joint average peak joint contact forces (×BW) and SDs during KW, WK, SW, and WS lifting conditions and 0°, 90°, and 180° turn of lift. Values correspond to averages shown in Figure 4A and 4B.

0° lifts ($P < .01$), and peak total forces were 18.1% lower in 180° versus 90° lifts ($P < .01$; Figure 4B, Table 1). Peak shear forces in the knee were 42.1% greater in descension versus ascension lifting tasks ($P < .01$; Figure 5B). At the knee, peak normal, shear, and total JCFs were 29.6%, 72.8%, and 32.0% higher in below-the-waist versus above-the-waist lifts ($P < .01$; Figure 6B).

There was a positive linear correlation ($R^2 = .60$, $P < .01$) between L5/S1 and knee and JCFs. That is, larger knee contact forces were associated with larger L5/S1 contact and vice versa (Figure 7).

Discussion

The purpose of this study was to go beyond limb joint analyses and establish a framework to examine how external loads resolve as 3-dimensional JCFs *inside limb joints* during MMH tasks. Specifically, to gain insight into potential for injury risk from exposure to acute peak loading during repetitive, asymmetric, and physically taxing movements,[4] we characterized both lower back (L5/S1) and knee joint loads in occupational MMH lifting tasks across 4 different starting positions and 3 degrees of twisting (Figure 1). We hypothesized that asymmetrical (H1), ascending (H2), and below the waist (H3) lifts would induce the highest JCFs at both the back and knee. Surprisingly, our data did not support H1 and H2 as asymmetric (180°) lifts had *lower* peak normal, shear, and total JCFs than 0° and 90° lifts (Figures 3 and 4) and lifting descension (WK/SW) tasks had *higher* peak shear JCFs only at the knee compared with ascension (KW/WS) tasks (Figures 3 and 5). In support of H3 and H4, we found that below-the-waist lifting tasks (KW/WK) had *higher* JCFs than above-the-waist (SW/WS) lifting tasks for both joints (Figure 6) and the knee and L5/S1 peak total JCFs had a positive linear correlation across the task space (Figure 7).

We anticipated asymmetrical lifting would generate more injurious JCFs than symmetrical lifting (H1) in the L5/S1 joint

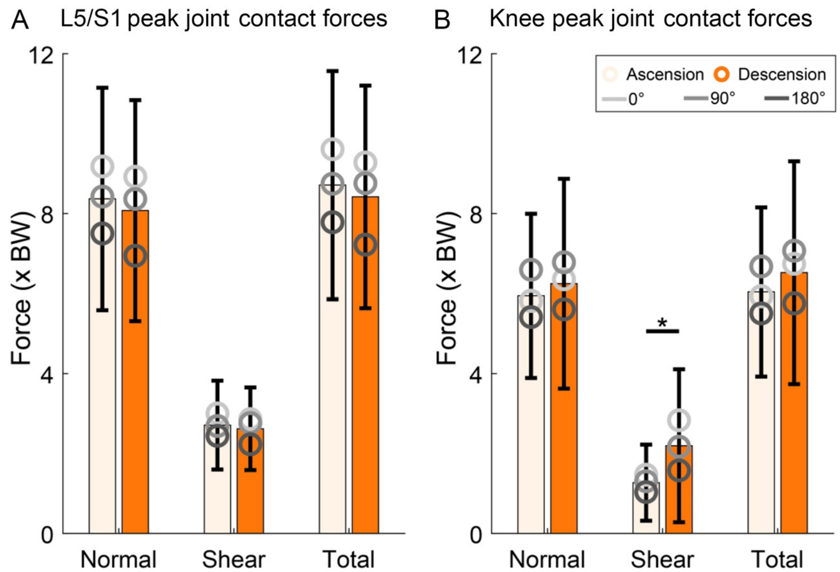

Figure 5 — The across-participant averaged peak normal, shear, and total JCFs in the L5/S1 joint (A) and knee (B) as explored in the second hypothesis (H2). The bars represent the JCF value averaged across the 3 lift degrees (0°, 90°, and 180°) for both the ascension (KW and WS) and descension (WK and SW) lifting tasks in orange. Independent values for each lift degree are labeled as circles in the corresponding color. Significantly different groups are denoted with an asterisk. Statistical significance across all conditions concluded when $\alpha = .05$. JCFs indicates joint contact forces; KW, knee-to-waist; SW, shoulder-to-waist; WK, waist-to-knee; WS, waist-to-shoulder.

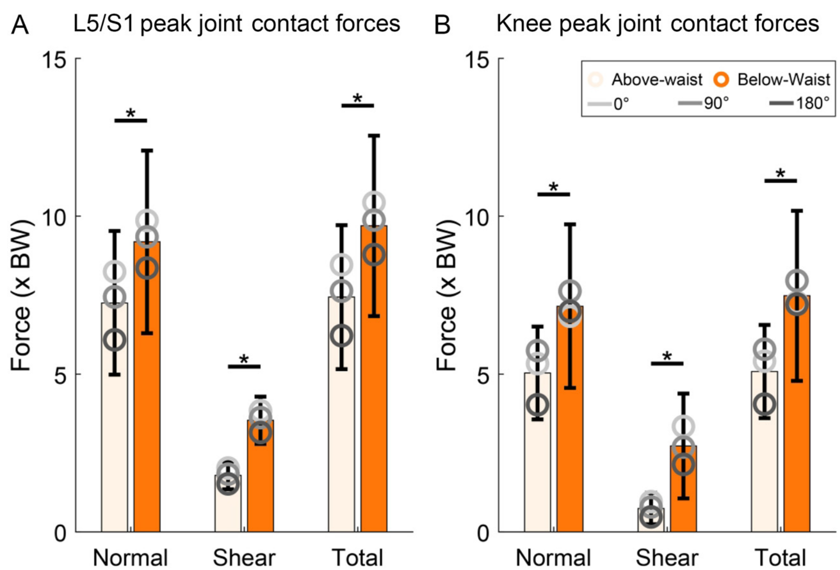

Figure 6 — The across-participant averaged peak normal, shear, and total JCFs in the L5/S1 joint (A), and knee (B), as explored in the third hypothesis (H3). The bars represent the JCF value averaged across the 3 lift degrees (0°, 90°, and 180°) for both the below-waist (KW and WK) and above-waist (SW and WS) lifting tasks in orange. Independent values for each lift degree are labeled as circles in the corresponding color. Significantly different groups are denoted with an asterisk. Statistical significance across all conditions concluded when $\alpha = .05$. JCFs indicates joint contact forces; KW, knee-to-waist; SW, shoulder-to-waist; WK, waist-to-knee; WS, waist-to-shoulder.

due to elevated shear forces caused by a shift in segmental centers of mass and increased external loading from the hands holding a weighted object.[11] However, our results suggest that *symmetric* lifting actually induced *higher* peak normal, shear, and total JCFs than *asymmetric* lifting across conditions (Figures 3 and 4A). Instances of greater lumbar flexion (Supplementary Figure S1B [available online]) aligned with increases in peak JCFs (Figure 3)—a postural shift that causes increased external loading and concomitant increases in muscle activations to stabilize the body, resulting in higher JCFs. Indeed, peak muscle activations, primarily from the erector spinae (ES), and peak L5/S1 joint moments in the sagittal plane were both greater in symmetric lifting than asymmetric lifting (Supplementary Figures S2 and S3 [available online]). In addition to these findings, we believe local peak JCFs are likely caused by large moment arms created through constraints within the execution of each lifting task. Participants in the study

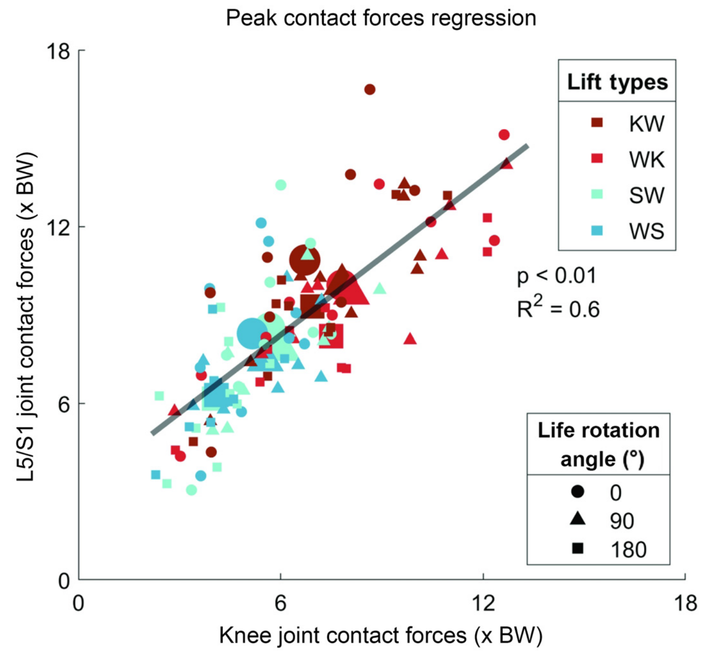

Figure 7 — Linear regression comparing peak total (net) contact forces between the L5/S1 and knee joints as explored in the fourth hypothesis (H4). All lift conditions (KW, WK, SW, and WS) and degrees of turn (0°, 90°, and 180°) combinations for all 9 participants are included in the regression analysis. Each participant's corresponding datapoints are represented by the smaller shapes, while the larger shapes represent across-participant averages for the conditions. JCFs indicates joint contact forces; KW, knee-to-waist; SW, shoulder-to-waist; WK, waist-to-knee; WS, waist-to-shoulder.

were constrained to place 1 foot on each force plate, configuring symmetric lifts side-by-side and asymmetric lifts diagonally adjacent. These stance configurations likely affected how they transported the weight from the initial to terminal shelf position. In 0° lifts, shelf placements caused participants to stand further away and extend the weight further from their center of mass. Thus, creating a larger moment arm about the L5/S1 joint, larger external moments about the lower back, and ultimately higher compressive and shear forces in the joint. In contrast, the asymmetric lifts (90° and 180°) did not present shelf obstructions, and participants could hold the weight closer to their torso throughout the lift. This technique yielded smaller moment arms between load and L5/S1 joint and resulted in lower compressive and shear forces in the joint.[12]

We predicted that ascending lifts would induce greater contact forces at the L5/S1 joint than descending lifts (H2) because we assumed the lumbar extension moment generated to counteract gravitational and inertial forces, both from the participant and the added mass being lifted, would be higher during ascending lifts[10] (ie, accelerating upward against gravity would be more demanding). However, we found no significant differences in joint loading when comparing ascension versus descension (Figure 5A). We found no significant difference in lifting durations between ascension (2.78 s) and descension (2.76 s). The weight lifted likely did not influence the rate of the lifts in our study or cause inertial effects as large as reported in other studies, where participants lifted up to 40 kg.[10,13]

We expected that below-the-waist lifting conditions (KW and WK) would cause higher JCFs than above-the-waist (SW and WS) lifting at the L5/S1 joint (H3) due to greater flexion in the trunk, knees, and hips.[40] In flexed postures, higher external moments are needed to compensate for poor mechanical advantage against body weight, shifts in centers of mass across segments,[11] and the load of the weighted object in participants' outstretched hands. Our results supported this rationale with below-waist lifts showing higher peak and integrated L5/S1 external moments (Supplementary Figures S2 and S3 [available online]) to go along with higher peak JCFs (Figures 3 and 6A). This confirms the intuition that lifts with higher degrees of trunk flexion places people in a vulnerable and weaker posture[48] that leads to higher shear forces[17] and increased injury risk.

We expected that adding asymmetrical twisting to a given lift would induce axial moments[49] and increase shear forces in the knee leading to higher JCFs when compared with symmetrical lifting (H1). Contrary to our expectation, 90° and 180° turns in lifting had an inconsistent effect on knee joint loading. We propose 2 potential explanations: (1) constrained foot placements altered participants' squatting stance width and/or (2) nonintuitive mechanics of the quadriceps in deep knee flexion. Escamilla et al[14] found knee (tibiofemoral) compressive forces 16% greater in wider stances than narrow stances. Following this line of reasoning, we saw that, on average, participants had wider stances and greater contact forces in 90° asymmetrical lifts (Figure 4B) compared with 0° and 180° lifts. On the other hand, Nisell and Ekholm[50] described a "wrapping effect" from the contraction of the quadriceps muscle pulling the patellar tendon posteriorly toward the femur (ie, intercondylar fossa) in very deep flexion, which can act to alleviate knee joint loading by providing an additional contact point for force distribution and transfer.[51] In addition, literature suggests peak knee compressive joint forces occur around 90° knee flexion in squatting.[51–53] This further supports our contention that the biomechanical technique that participants used when lifting symmetrically resulted in deeper knee flexion and incited the indirect benefits of the wrapping effect to reduce peak contact forces.

During squat descension, the knee extensor moment acts to control deceleration to levels below that induced by gravitational forces. In theory, this external mechanical demand should be less than that in ascent, when the extension moment must first overcome gravitational forces before it can accelerate the limbs—and as a result, we expected JCFs to be lower in descent versus ascent (H2). Surprisingly, we found little difference in JCFs between ascending and descending lifts at the knee. *Only* shear forces showed differences, with statistically greater contact forces appearing in descension versus ascension (Figure 5B), potentially due to the inertial effects of the added mass being lifted. These results agree with the findings of Dahlkvist et al[54] who also observed similarities in knee joint forces between squats in descent and ascent. De Looze et al[15] explain that similar muscle activity in ascent and descent lead to loading similarities but suggest that descending tasks are more predisposed to injury occurrence due to muscle lengthening during loading. Similarly, Van Rossom et al[55] found that medial knee (tibiofemoral) peak and average shear forces are greater in stair descent than in stair ascent, which further solidifies the idea that the momentum of deceleration leads to greater JCFs. Taken together, we interpret these studies to suggest that lifting may be safer than lowering heavy weight if the goal is to reduce joint loading on the knees.

Our results support our hypothesis that below-the-waist lifting (KW and WK) would generate higher contact forces than above-the-waist (SW and WS) lifting for knee (H3; Figure 6B). Delisle et al[56] found that minor knee flexion alleviates some stress from the L5/S1 joint in lifting. Indeed, we observed that above-the-waist lifting conditions in this study were performed with only slight knee flexion compared to below-the-waist lifts. In a nearly upright posture, activations and forces of the quadriceps (knee extensors), hamstrings, and gastrocnemius (knee flexor) are known to be relatively quiescent,[54] yielding smaller knee JCFs than in below-the-waist lifting.

In this study, we allowed participants to lift without a predefined technique to assess the effect of natural repetitive lifting on internal joint loading and participants seemed to prefer movements that traded-off loads in lower back and knee. We found that, across all lifting tasks, peak JCFs between the L5/S1 and knee joints scaled linearly (Figure 7) perhaps reflecting a motor control principle to limit peak loading at any one joint. Studies have confirmed that posture is an important factor,[10,40] which can influence load distribution across the body or off-loading away from a joint[49] in lifting tasks. Lifting techniques, such as stoop (back lift) versus squat (leg lifting),[13,15,48,57–59] induce postural constraints that can bias applied external loads and joint moments.[49] For example, studies have shown that lifting form is user-specific and is dependent on the starting height of the movement, but also the trunk, knees, and hips become more kinematically coordinated in below-waist lifting; hence why often the preferred lifting technique is a blend between stooping and squatting.[9,41] Along these lines, Splittstoesser et al[60] conducted a lifting study while kneeling, fixing the role of the knee joint. Interestingly, they found that without contributions from the knee to perform the lift, moving loads to positions above the waist led to greater compressive forces in the back. Thus, highlighting the potential importance of coordinating lower limb joints to reduce loading in the lower back.[40] Indeed, Bejjani et al[42] concluded that both the back and the knees must bend about 60° and 90°, respectively, to reduce the average force distributed between both joints; thus posing the question if individuals might naturally opt for lifting postures which produce lower average JCFs.

There are several limitations which could have affected our experimental outcomes. We controlled the weight being lifted (11.34 kg), lifting conditions (starting-ending positions and degree of turn), and foot placement; however, the rate of lifting, consistency in reach distance from shelf, and stance width are all factors which were loosely regulated and may have contributed to variance in our results. We collected EMG from 14 muscles across the torso and legs, limiting CEINMS to simulate most muscle activity. Knee JCF results were focused on the right, EMG-instrumented leg, but we expect similar forces on the left knee. Finally, the lack of female representation in this study restricted us from investigating the effects of sex on JCFs.

This study investigated how JCFs in the knee and lower back (L5/S1) vary during manual lifting tasks. We found highest JCFs in symmetric (ie, load straight in front of the body), below the waist lifting tasks. Taken together, our results suggest that asymmetric lifting can *actually reduce* JCFs in the lower back and knee when compared to moving loads straight in front of the body. Interventions to reduce JCFs during heavy lifting should emphasize transporting weight above the waist or close to the body, while encouraging postures that minimize trunk and knee flexion (ie, small external moment arms), but to still use caution in asymmetric postures. Finally, we found that JCFs were proportionally distributed across knee and lower back joints, suggesting that preferred lifting techniques might be selected to balance JCFs across joints and avoid local overloading—a motor control principle indicating a trade-off that might be leveraged by assistive technology applied at 1 joint (eg, back) to protect another (eg, knee).

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