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Unilateral transtibial prosthesis users load their intact limb more than their prosthetic limb during sit-to-stand, squatting, and lifting

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ABSTRACT

Background: Lower limb prosthesis users exhibit high rates of joint pain and disease, such as osteoarthritis, in their intact limb. Overloading of their intact limb during daily activities may be a contributing factor. Limb loading biomechanics have been extensively studied during walking, but fewer investigations into limb loading during other functional movements exist. The purpose of this study was to characterize the lower limb loading of transtibial prosthesis users during three common daily tasks: sit-to-stand, squatting, and lifting.
Methods: Eight unilateral transtibial prosthesis users performed sit-to-stand (from three chair heights), squatting, and lifting a 10 kg box. Peak vertical ground reaction forces and peak knee flexion moments were computed for each limb (intact and prosthetic) to characterize limb loading and asymmetry. Ranges of motion of the intact and prosthetic ankles were also quantified.
Findings: Users had greater peak ground reaction forces and knee flexion moments in their intact limb for all tasks ($p < 0.02$). On average, the intact limb had 36–48% greater peak ground reaction forces and 168–343% greater peak knee flexion moments compared to the prosthetic limb. The prosthetic ankle provided <10° of ankle range of motion for all tasks, less than half the range of motion provided by the intact ankle.
Interpretation: Prosthesis users overloaded their intact limb during all tasks. This asymmetric loading may lead to an accumulation of damage to the intact limb joints, such as the knee, and may contribute to the development of osteoarthritis. Prosthetic design and rehabilitation interventions that promote more symmetric loading should be investigated for these tasks.

1. Introduction

Lower limb prosthesis users are a diverse population encompassing individuals of all ages, activity levels, and lifestyles and with varying causes of limb loss. However, there are significant impacts of being a prosthesis user that are prevalent throughout the population including challenges to mobility, quality-of-life, and long-term health.

Many unilateral lower limb prosthesis users (LLPUs) develop secondary physical conditions, such as low back pain, osteoarthritis, and osteoporosis (Gailey, 2008). Specifically, they are at greater risk of musculoskeletal injury, joint degeneration and pain in their intact (non-prosthetic) limb compared to the general population (Gailey, 2008). Previous observational studies have reported unilateral LLPUs are more likely to develop knee and hip osteoarthritis and have a high incidence of pain in the joints of their intact limb. The prevalence of osteoarthritis in LLPUs is broadly estimated to be between 16 and 60% for the knee (Lemaire and Fisher, 1994; Melzer et al., 2001; Norvell et al., 2005; Struyf et al., 2009; Wasser et al., 2022) and 14% for the hip (Struyf et al., 2009) compared to 2–11% of the general population experiencing knee or hip osteoarthritis (Norvell et al., 2005; Struyf et al., 2009). Intact limb joint pain is also a primary concern for long-term prosthesis users, with over half (55%) reporting knee pain and 23% reporting hip pain (Mussman et al., 1983). These findings of increased joint pain and osteoarthritis among LLPUs remain even when age, gender, and body mass are controlled (Melzer et al., 2001; Norvell et al., 2005).

Previous work has proposed the increased pain and injury to the intact limb potentially comes from LLPUs employing altered movement strategies during daily tasks that overload their intact limb relative to

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Table 1
Individual and mean (standard deviation) participant demographics for unilateral transtibial prosthesis users.

Participant ID	Sex	Age (years)	Mass (kg)	Height (m)	Years since Limb Loss	Prosthetic side	K-Level	Cause of Limb Loss	Daily Use Prosthesis
1	M	30	86.9	1.77	7.5	Left	4	Traumatic	Fillauer Formula
2	M	26	76.2	1.74	8.5	Left	4	Traumatic	Fillauer AllPro
3	M	52	102.8	1.87	10.2	Left	4	Traumatic	Fillauer AllPro
4	M	32	88.5	1.86	2	Left	4	Traumatic	Blatchford Echelon
5	M	40	69.5	1.76	9	Left	4	Traumatic	Össur Vari-Flex
6	F	42	68.6	1.53	4	Left	4	Traumatic	Össur Pro-Flex Pivot
7	M	66	107.0	1.78	21	Right	4	Traumatic	Blatchford Elan
8	M	27	86.6	1.97	0.5	Right	4	Traumatic	Fillauer AllPro
Mean ± SD		39.4 ± 13.9	85.8 ± 14.1	1.78 ± 0.13	7.8 ± 6.4				

their prosthetic limb (Butowicz et al., 2017; Gailey, 2008; Miramand et al., 2022). During walking, LLPUs often exhibit asymmetrical motion (e.g., spatiotemporal parameters, joint kinematics), joint moments, and ground reaction forces, where the intact limb bears more load than the prosthetic limb (Sagawa et al., 2011). LLPUs have increased intact limb stance time indicating they spend more time on their intact limb than prosthetic limb while walking (Nolan et al., 2003; Sanderson and Martin, 1997). When comparing peak vertical ground reaction forces during walking, the intact limb experiences up to 21% more force than the prosthetic limb depending on the type of prosthesis used (Engsberg et al., 1993; Menard et al., 1992; Nolan et al., 2003; Powers et al., 1994; Snyder et al., 1995). In contrast, individuals without limb loss display <10% asymmetry in peak force between limbs during walking (Herzog et al., 1989; Menard et al., 1992; Nolan et al., 2003).

Repeated elevated mechanical loading on the intact limb joints can cause tissue damage to accumulate over time and is concerning as long-term exposure can lead to the degeneration of weight-bearing joints and joint pain (Amiri et al., 2023; Griffin and Guilak, 2005; Radin and Rose, 1986). While the greater susceptibility of LLPUs to secondary musculoskeletal conditions such as osteoarthritis is multifactorial, the increased loads experienced by the joints of the intact limb over long periods of time is likely a primary contributor.

While there exists considerable investigation into LLPU performance and lower limb loading during walking, less work has investigated user performance during other activities of daily living. Activities such as standing up from a chair, picking up an item off the floor, and squatting down are everyday movements that are essential for independent living, but are challenging tasks for many LLPUs (Devan et al., 2015; Gauthier-Gagnon et al., 1999; Hafner et al., 2016; Waldera et al., 2013). Sit-to-stand is an important and biomechanically demanding task that LLPUs and individuals without limb loss do over 50 times per day (Bussmann et al., 2008). While some LLPUs struggle to stand up from a chair independently, those who can have been shown to complete the sit-to-stand motion asymmetrically by having greater ground reaction forces under their intact limb compared to their prosthetic limb (Miramand et al., 2022). Increased intact limb joint moments and lumbar loads compared to individuals without limb loss have also been observed (Actis et al., 2018; Agrawal et al., 2011; Miramand et al., 2022; Šlajpah et al., 2013).

The general motion of squatting down or lifting an object is also common in daily life and often required for recreation, exercise, and in some occupations. Squatting and lifting involve lowering and raising one's center-of-mass through coordinated flexion and extension of the lower limb joints. A proportion of LLPUs cannot squat, and even those with high levels of functional ability often find it difficult (Devan et al., 2015; Gauthier-Gagnon et al., 1999; Morgan et al., 2022; Waldera et al., 2013). Lifting has been studied extensively in individuals without limb loss as jobs that involve strenuous lifting have high rates of musculoskeletal injury. However, there exists little investigation into how LLPUs perform squatting or lifting. The limited existing work focuses on low back muscle activation in occupational lifting tasks (Fu et al., 2015; Jayaraman et al., 2020; Lieber et al., 2002) or presents case studies of one LLPU performing a weighted back squat (Cooper et al., 2022; Mrazsko et al., 2020). While lifting and squatting are common daily tasks performed frequently by LLPUs, there exists no characterization of the lower limb loading of this population during these tasks.

The objective of this study was to characterize the lower limb loading of unilateral transtibial prosthesis users during sit-to-stand, squatting, and lifting. We hypothesized LLPUs would load their intact limb more than their prosthetic limb, as indicated by greater peak ground reaction forces and knee flexion moments in the intact limb. We also explore user and prosthesis performance to gain insight into why users have challenges with these movements and elucidate directions for future interventions.

2. Methods

We analyzed the lower limb loading of LLPUs completing sit-to-stand, squatting, and lifting while wearing their clinically prescribed prosthesis. Participants completed these tasks as part of a larger data collection that also included reaching and lunging. Through reviewing the scientific literature and interviewing local LLPUs, physicians, and prosthetists, we identified this larger set as common daily tasks that LLPUs often avoid or find challenging. In this study, we focused our analysis on sit-to-stand, squatting, and lifting. These are bilateral movements in which individuals without limb loss typically move and load their lower limbs symmetrically. Lunging and reaching will form the basis of a separate study due to the more asymmetrical nature of these movements.

2.1. Participants

A convenience sample of eight unilateral transtibial LLPUs (7 M/1 F, age: 39.4 ± 13.9 years, mass: 85.8 ± 14.1 kg, height: 1.78 ± 0.13 m) was recruited for this study (Table 1). Participants were included if they were at least four months post-surgery, could walk without an aid, and had no recent injuries that impacted mobility. All participants provided written informed consent according to Vanderbilt Institutional Review Board approved procedures.

2.2. Experimental protocol

Prosthesis users wore their prescribed prosthesis (Table 1) and their preferred footwear during data collection. Ground reaction force (GRF) data were recorded under each foot at 1000 Hz using in-ground force plates (AMTI, Watertown, MA, USA) and lower-body kinematics were recorded at 200 Hz using a 10-camera motion capture system (Vicon, Oxford, UK). Passive reflective markers were affixed to the trunk (6), pelvis (6), thighs (8), knees (4), and shanks (8). On the intact limb, markers were placed on the medial and lateral malleoli (2), calcaneus (3), first and fifth metatarsal base and head (4), and the second toe (1). On the prosthetic limb, the medial and lateral malleoli markers were placed on the prosthesis at the same height as the malleoli markers on the intact limb. The prosthetic foot markers were mirrored from the shoe of the intact limb.

Participants were instructed to complete the tasks in the way that felt most comfortable to them. They were not instructed to complete the tasks at any specific pace or with any specific movement pattern. Foot position was not controlled besides the constraint that each foot had to remain on a separate force plate. If a participant could not complete a task, this was documented before proceeding to the next task. For sit-to-stand, data were collected as the participant stood up from a chair of standard height (48 cm) that had a backrest, no armrests, and no padding (Chorin et al., 2015; Schofield et al., 2013). Two more variations of the sit-to-stand task were performed at lower chair heights of 38 cm and 28 cm. Standing up from a low chair was included in this protocol because it was specifically mentioned as being difficult by prosthesis users and clinicians during interviews. Participants did not push on the chair with their arms to assist them in standing up. For squatting, participants started from standing with a foot under each shoulder and were instructed to "squat down to a comfortable level" before returning to standing. The lifting task required participants to pick up a 10 kg box ($27 \times 27 \times 40$ cm) from the floor in front of them and return to standing. The parameters of this task were chosen to provide a simple and reasonable representation of lifts encountered in daily life. Each task was repeated three times (Chorin et al., 2015) resulting in fifteen total trials. Tasks were completed in the order listed and breaks between tasks were provided as needed.

After each task, participants were asked to verbally rate effort, stability, and comfort on a scale of 0 to 10. For perceived effort, a rating of 0 indicated "the task took no effort" and a rating of 10 indicated "the task took maximum effort." For stability and comfort, a rating of 0 indicated the participant felt "completely unstable" or "completely uncomfortable" and a rating of 10 indicated the participant felt "completely stable" or "completely comfortable" doing the task.

2.3. Data processing and analysis

We computed two complementary metrics to characterize the lower limb loading of participants: vertical ground reaction forces (GRF) and knee flexion moments. The vertical GRF is the primary direction of the force under the feet during these tasks (Chorin et al., 2015), and it is commonly used as an indicator of the overall loading strategy of LLPUs during activities of daily living (Agrawal et al., 2011; Simon et al., 2016; Wolf et al., 2013). Large asymmetries in GRF during a bilateral task indicate participants are bearing more weight on one limb over the other, which often contributes to increased moments and contact forces in the joints of the overloaded limb (Ding et al., 2023; Ferris et al., 2017; Nolasco et al., 2020; Šlajpah et al., 2013). Knee moments have been used as an indicator of loading of the internal structures of the knee joint (Creaby, 2015; Morgenroth et al., 2012; Wasser et al., 2020; Zeighami et al., 2021). The sagittal plane knee moment, often reported as the external knee flexion moment, has been shown to correlate to both modeled and direct measurements of forces within the knee (D'Lima et al., 2007; Trepczynski et al., 2014). Especially for bilateral tasks that require a high degree of knee flexion (e.g., sit-to-stand, squatting), an increase in knee flexion moment is indicative of an increase in knee joint contact forces both in the tibiofemoral and patellofemoral spaces (Ding et al., 2023; Nagura et al., 2002; Trepczynski et al., 2014).

We also computed the range of motion (RoM) of the intact and prosthetic ankle during each task. Ankle RoM was computed as a secondary metric to characterize the performance of the prostheses during these tasks. For individuals without limb loss, sit-to-stand, squatting, and lifting typically involve a large amount of ankle flexion (Grimmer et al., 2020), which most ankle-foot prostheses lack. Characterizing the prosthesis RoM could help explain user challenges, limb loading asymmetries, and elucidate directions for future prosthetic interventions.

GRF data and marker trajectories were low-pass filtered at 15 Hz and 8 Hz, respectively, before analysis. The start and end of each task repetition was determined using the vertical movement of the marker placed on the C7 vertebrae of participants (with approximately 0.5 cm of increase or decrease in position used as a threshold of movement). For sit-to-stand, only the standing up portion of the movement is included in this analysis. This task started when the participant began moving from seated and ended when they were standing. The squatting task started when the participant began to lower and ended when they returned to standing upright. Similarly, lifting began when the participant started bending down and ended when they returned to standing holding the 10 kg box.

Peak vertical GRF under each limb was computed for each repetition before being averaged across the three repetitions of each task. For a small number of lifting trials, some participants shifted their weight from one foot to the other as they set up to complete the task, which resulted in a brief peak in GRF. In these instances, force data were manually cropped to ensure that peak GRFs were only extracted while the task was actually being completed. GRF data were normalized by participant body weight. We used Visual3D (C-motion, Germantown, MD, USA) to create an eight-segment model (trunk, pelvis, two thighs, two shanks, and two feet). Segment inertial properties were set to the default values built into the software. Inertial properties of the prosthetic limb were assumed to be identical to the intact limb as previous work has observed this has minimal impact on knee moments during stance (Reinbolt et al., 2007). We used this model and inverse dynamics to compute the net external knee joint moment in the sagittal plane. We report this as the knee flexion moment with flexion defined as positive. Peak knee flexion moment was computed for each task repetition then averaged across repetitions. Moment data were normalized by participant body weight and height before being averaged across participants.

Ankle angle was computed in Visual3D as the angle between the shank segment and the rearfoot segment in the sagittal plane for each lower limb. The motion of the rearfoot segment was tracked by the movement of three markers on the calcaneus for almost all trials. For a small number of sit-to-stand trials from the lower chairs ($N = 5$), a calcaneus marker was occluded by the chair so the model of the rearfoot was altered to include the markers on the base of the first and fifth metatarsals in addition to the calcaneus markers. A sensitivity analysis revealed the effect of using these alternate markers was negligible ($<1^\circ$ change in results). Ankle RoM for each limb was computed as the difference between the minimum and maximum ankle angle. This was computed for each repetition before averaging across the three repetitions for each task.

Descriptive statistics (mean $\pm$ standard deviation) were computed, and the data were screened for normality via a Shapiro-Wilk test. Following this, differences in peak vertical GRF, peak knee flexion moment, and ankle RoM between the intact and prosthetic limb for each task were assessed using a paired t-test (normally distributed) or Wilcoxon signed-rank test (non-normally distributed) with an alpha level of 0.05.

2.4. Control data

Eight individuals without limb loss were recruited as control participants (5 M/3 F, age: 35.9 ± 9.7 years, mass: 78.4 ± 14.6 kg, height: 1.77 ± 0.09 m). They were tested with the same protocol to provide a reference for interpretation of LLPU results (participant details in Table S1). This dataset provides a comparison for the exact tasks performed by the LLPU participants in this study. For control participants, peak GRF, peak knee flexion moment, and ankle RoM were compared between their dominant and non-dominant limb. The dominant limb was defined as the self-reported leg they use to kick a ball (van Melick et al., 2017).

3. Results

Our core analysis focuses on sit-to-stand from a standard chair height (48 cm), squatting, and lifting a 10 kg box. All participants were able to complete three repetitions of these three core tasks. A brief analysis comparing sit-to-stand from three chair heights (48 cm, 38 cm, 28 cm) is

Table 2
Mean (standard deviation) peak vertical ground reaction force, peak knee flexion moment, and ankle range of motion for the intact and prosthetic limb of LLPUs. Force was normalized to body weight (BW) and moment was normalized to body weight and height (Ht). A significant difference was detected between limbs for all metrics and all tasks.

	Peak Vertical Ground Reaction Force (BW)			Peak Knee Flexion Moment (BW*Ht)			Ankle Range of Motion (°)		
	Intact Limb	Prosthetic Limb	*P*-value	Intact Limb	Prosthetic Limb	P-value	Intact Ankle	Prosthetic Ankle	P-value
Sit-to-Stand (Standard, 48 cm)	0.77 ± 0.08	0.56 ± 0.05	0.001	0.074 ± 0.018	0.017 ± 0.005	0.008	13.4 ± 3.0	6.2 ± 1.9	<0.001
Squat	0.81 ± 0.09	0.58 ± 0.11	0.008	0.079 ± 0.018	0.025 ± 0.010	0.008	25.5 ± 5.9	6.6 ± 3.0	<0.001
Lift	0.85 ± 0.09	0.61 ± 0.05	0.008	0.051 ± 0.023	0.019 ± 0.001	<0.001	20.8 ± 7.3	7.8 ± 2.8	<0.001
Sit-to-Stand (Low, 38 cm)	0.82 ± 0.07	0.57 ± 0.06	<0.001	0.084 ± 0.014	0.021 ± 0.006	0.008	18.2 ± 3.8	7.5 ± 2.7	0.008
Sit-to-Stand (Lowest, 28 cm)	0.84 ± 0.05	0.57 ± 0.06	<0.001	0.095 ± 0.011	0.022 ± 0.009	0.016	21.1 ± 4.2	7.2 ± 1.4	<0.001

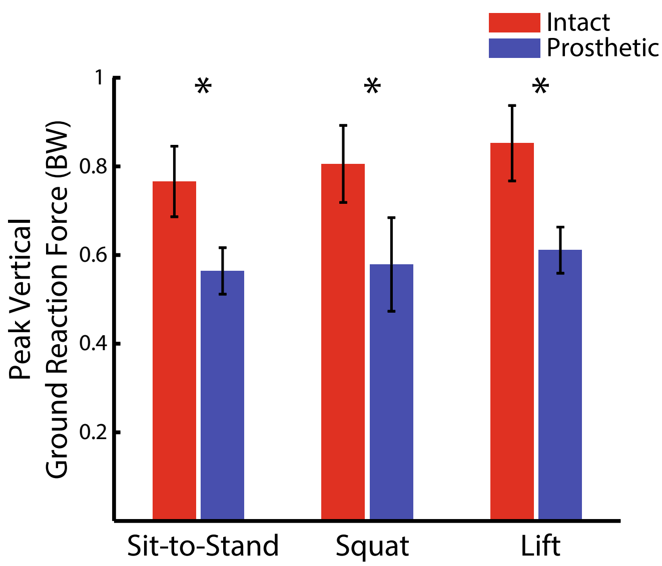

Fig. 1. Peak vertical ground reaction force under each limb during sit-to-stand (48 cm), squatting, and lifting ($N = 8$). Force is normalized to participant body weight (BW). *A significant difference was detected between the intact limb (red) and the prosthetic limb (blue) for all tasks. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

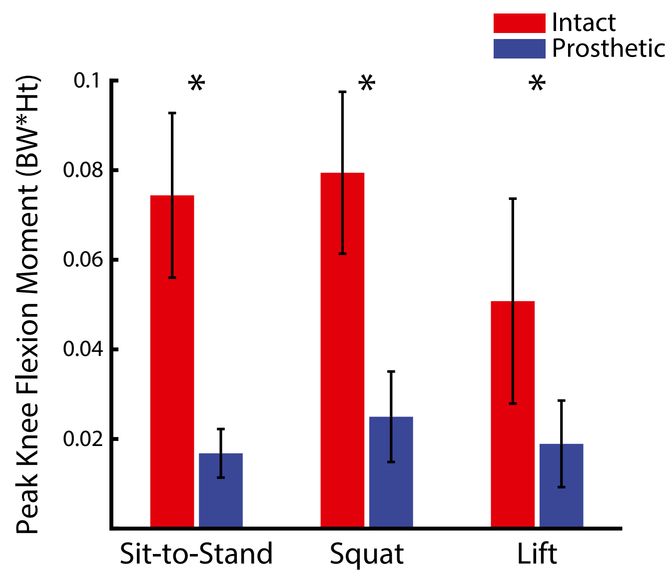

Fig. 2. Peak knee flexion moments during sit-to-stand (48 cm), squatting, and lifting (N = 8). Moment is scaled by participant body weight (BW) and height (Ht). *A significant difference was detected between the intact limb (red) and the prosthetic limb (blue) for all tasks. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

also included at the end of this section.

3.1. Ground reaction force and knee flexion moment

Peak GRF and peak knee flexion moment were significantly greater in the intact limb compared to the prosthetic limb during sit-to-stand from a standard chair height, squatting, and lifting a 10 kg box ($p <$ 0.01 for all three tasks, Table 2). On average, peak vertical GRFs for the intact limb were greater than the prosthetic limb by 36% during sit-to-stand, 39% during squatting, and 39% during lifting (Fig. 1, Table 2). Peak knee flexion moments in the intact limb were greater than the prosthetic limb by 343% during sit-to-stand, 218% during squatting and, 168% during lifting (Fig. 2, Table 2). For control participants, peak GRFs between the dominant and non-dominant limb were typically within 5% ($p >$ 0.1; Fig. S1, Table S2) and peak knee flexion moments between limbs were within 3% ($p >$ 0.4; Fig. S2, Table S2).

Representative time-series plots and individual participant results for peak GRF and peak knee flexion moment can be seen in Figs. 3 and 4, respectively. All participants overloaded their intact limb for all three tasks. Every participant had a greater peak GRF (Fig. 3) and peak knee flexion moment (Fig. 4) in their intact limb relative to their prosthetic limb. Most participants continually put more force through their intact limb and had an increased knee flexion moment throughout the duration of each task, not just at the instance of peak force or moment. Corresponding time-series and participant-specific plots for control participants are provided in the Supplementary Materials (Figs. S3 and S4).

3.2. Ankle range of motion

On average, the intact ankle RoM of LLPUs was 13°, 25°, and 21° during sit-to-stand, squatting, and lifting, respectively. This is significantly more RoM than the prosthetic ankle provided which was only 6–8° for these tasks ($p < 0.001$; Fig. 5, Table 2). There was no significant difference in ankle RoM between the limbs of control participants ($p >$ 0.08). Their ankle RoM was approximately 19°, 32°, and 33° during sit-to-stand, squatting, and lifting, respectively (Fig. S5, Table S2).

3.3. Participant ratings of task difficulty

When rating their perceived effort on a scale of 0 to 10, participants reported sit-to-stand from a standard chair height, squatting, and lifting a 10 kg box required little effort indicated by a maximum rating of 2 (with 0 indicating "the task took no effort"). The exception to this was P7, the oldest participant (66 years old), who gave sit-to-stand from a standard chair height an effort rating of 3 and lifting a rating of 6. Participant ratings of stability and comfort tracked with their ratings of effort indicating they also felt stable and comfortable during these tasks.

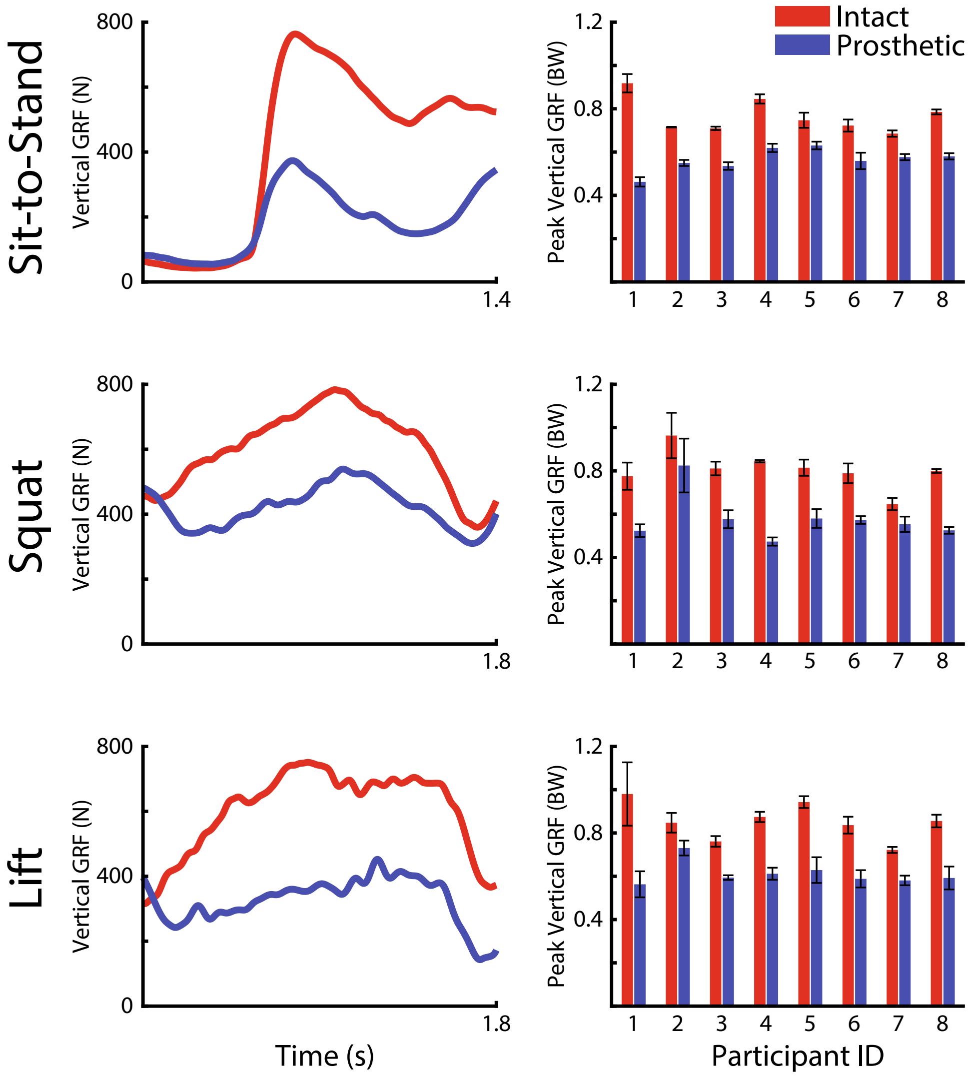

Fig. 3. Left: Representative time-series data of the vertical ground reaction force (GRF) during sit-to-stand (48 cm), squatting, and lifting. Right: Participant-specific averages of peak vertical ground reaction force during each task. Peak force values were normalized to bodyweight (BW).

All stability and comfort ratings were >7 (with 10 indicating they felt "completely stable" or "completely comfortable"). All participants' individual ratings are provided in the Supplementary Material (Table S3).

3.4. Sit-to-stand across chair heights

When testing sit-to-stand from the two lower chair heights, all participants were able to complete these tasks except one. Participant 7 was unable to stand up out of the lowest chair (28 cm) so his biomechanics data for that task were not included in analysis. While participants reported standing up from a standard chair height as taking little effort, standing up out of the lowest chair was reported as challenging (Table S3). The majority of LLPUs gave the effort required to stand up from the lowest chair a score of 5 or more on the scale of 0 to 10 (with 10 being "the task took maximum effort"). Participants also reported feeling unstable and uncomfortable while standing up out of the lowest chair.

The significant difference in GRF, knee flexion moment, and ankle RoM observed during sit-to-stand from a standard height chair was also observed for the two lower chairs tested (Fig. 6, Table 2). For the low chair (38 cm) and the lowest chair (28 cm), LLPUs put 44% and 48% more force through their intact limb ($p < 0.001$). Peak knee flexion moments were 298% and 324% greater in the intact limb for the low and lowest chairs ($p < 0.02$). The average RoM of the intact ankle during these two tasks was 18° and 21° while only 7° of RoM was provided by the prosthetic ankle ($p < 0.01$). Control participants had no significant difference in peak GRF, peak knee flexion moment, or ankle RoM between limbs at any chair height ($p > 0.4$; Fig. S6, Table S2).

4. Discussion

Unilateral transtibial LLPUs asymmetrically load their lower limbs during sit-to-stand, squatting, and lifting. Users had greater peak GRFs (Fig. 1) and peak knee flexion moments (Fig. 2) in their intact limb compared to their prosthetic limb. This overloading of the intact limb was seen for every study participant and across the duration of the tasks tested (Figs. 3-4). During these tasks, the RoM of the prosthetic ankle was significantly less than the intact ankle for LLPUs (Fig. 5). The greater peak GRF, peak knee flexion moment, and ankle RoM in the intact limb observed during sit-to-stand from a standard chair height was also observed during sit-to-stand from two lower chairs (Fig. 6).

The current study confirms prior experiments that measured increased vertical GRFs and knee flexion moments in the intact limb compared to the prosthetic limb of LLPUs during sit-to-stand (Agrawal et al., 2011; Ferris et al., 2017; Nolasco et al., 2020; Šlajpah et al., 2013) and builds upon this prior analysis by showing these loading

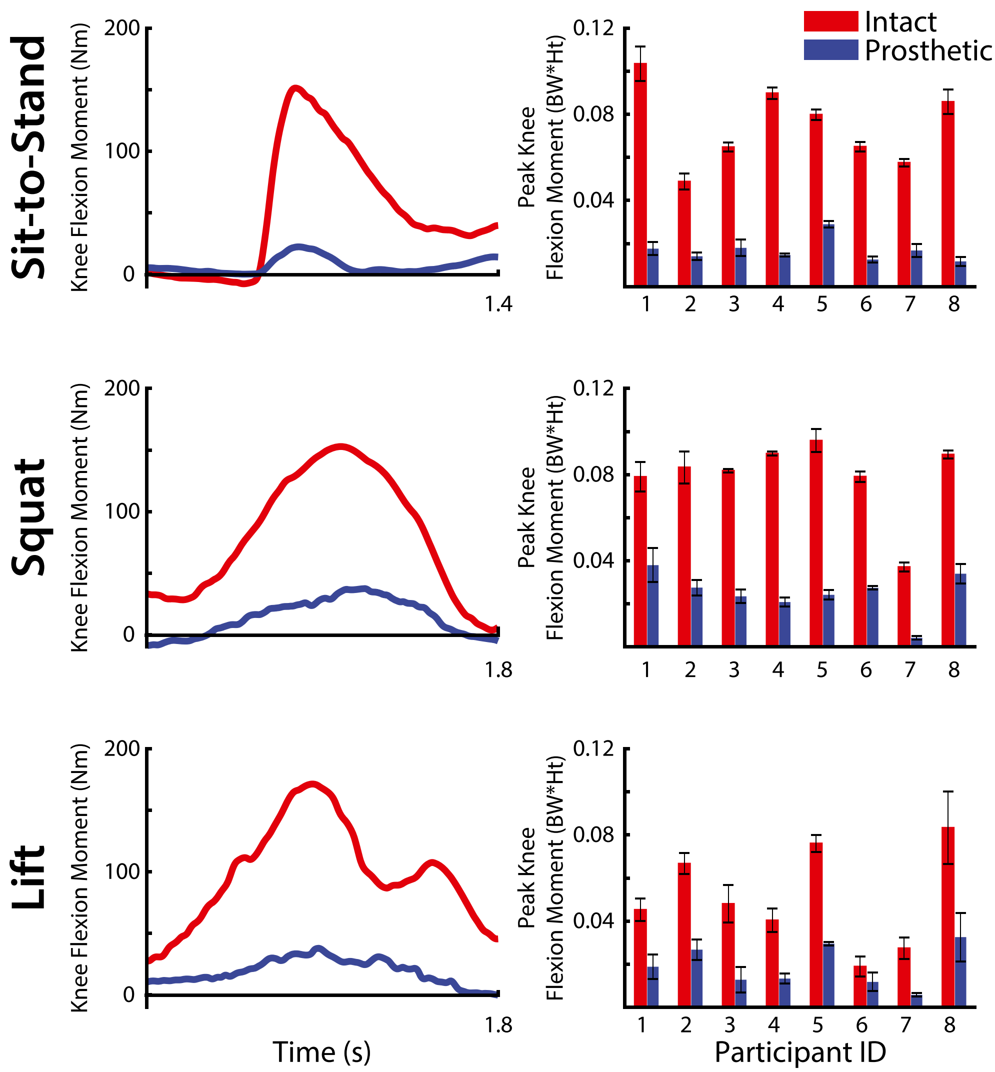

Fig. 4. Left: Representative time-series data of the knee flexion moment during sit-to-stand (48 cm), squatting, and lifting. Right: Participant-specific averages of peak knee flexion moment during each task. Peak moment values were normalized to bodyweight (BW) and height (Ht).

asymmetries also exist during squatting and lifting. This is the first study, to our knowledge, characterizing the lower limb loading of a group of LLPUs during squatting and lifting which are essential daily tasks. In addition to asymmetrical loading, the intact limb of LLPUs generally exhibited higher peak GRF and knee flexion moments than those experienced by either limb of control participants (Tables 2 & S2). Overloading the intact limb repeatedly during these common daily movements could result in joint damage accumulation and contribute to the development of joint degeneration and pain over time (Radin and Rose, 1986).

The increased load LLPUs put on their intact limb and specifically their knee joint during daily tasks may contribute to the high rates of joint degeneration and pain observed in the LLPU population. Previous research in walking and other tasks has shown increases in GRFs and knee moments correlate to increased knee contact forces, loads on the cartilage in the knee, and prevalence of cartilage degeneration (Ding et al., 2023; Lynn et al., 2007; Morgenroth et al., 2014; Wasser et al., 2020). Observational studies have suggested that high knee loads are a risk for the initiation and progression of osteoarthritis (Amiri et al., 2023; Chehab et al., 2014; Teng et al., 2017). These findings support the association between the mechanical loading of the knee joint and the development of osteoarthritis. Overloading of the intact limb during walking has been proposed to be a contributor to the high rates of joint pain and osteoarthritis observed in LLPUs. This study highlights the overloading of the intact limb during other common daily movements, which could be another contributor to the increased mechanical loading of the intact limb joints and the resulting joint degeneration and pain experienced by many LLPUs.

The results of this study indicate that LLPUs are overloading their intact limb, even during tasks they perceive as easy and comfortable. Thus, they may be accumulating damage to the joints of their intact limb without realizing it. The LLPUs in this study all have a high level of functional ability (all K4) and on average, are relatively young (39.4 ± 13.9 years). All participants were able to complete all tasks included in this analysis (except P7, who was unable to stand up from the lowest chair) with minimal effort (excluding standing up from a low chair). Sit-to-stand, squatting, and lifting are tasks known to be challenging for many LLPUs with a significant number of users unable to stand up from a chair independently or pick up an item off the floor (Gauthier-Gagnon et al., 1999). Even though the users in this study did not find most tasks challenging, on average they had 36–48% greater peak GRFs and 168–343% greater peak knee flexion moments in their intact limb compared to their prosthetic limb. Additionally, because the LLPUs in this study are relatively young, they will likely be prosthesis users for decades. Frequent overloading of the intact limb during daily tasks throughout their lifetime may further increase their risk of intact limb joint pain, knee osteoarthritis, and other musculoskeletal injuries.

Although LLPUs are a heterogenous population, often exhibiting

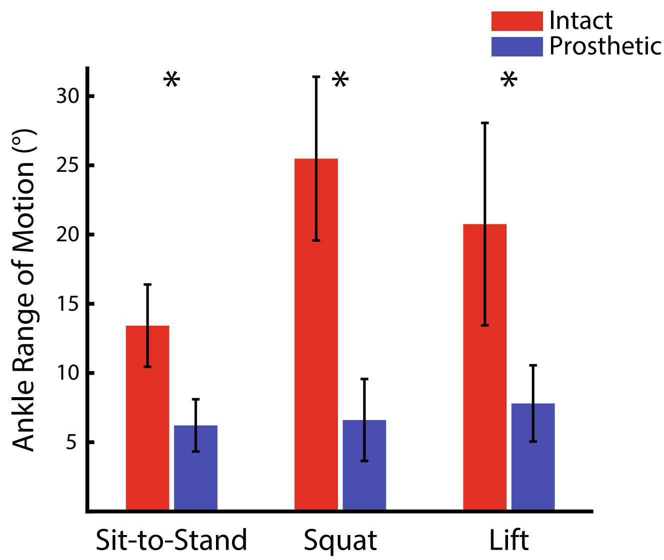

Fig. 5. Ankle range of motion during sit-to-stand (48 cm), squatting, and lifting ($N = 8$). *A significant difference was detected between the intact limb (red) and the prosthetic limb (blue) for all tasks. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

individualized movement strategies (Lamers et al., 2019; McDonald et al., 2022), the present study found that overloading of the intact limb during daily tasks was ubiquitous for all eight participants (Figs. 3-4). These participants had similar functional ability, but used various prescribed prostheses and differed in age, mass, height, and time using a prosthesis (Table 1). Despite these differences, all participants relied more on their intact limb than their prosthetic limb when completing sit-to-stand, squatting, and lifting. The ubiquitous overloading of the intact limb observed in this study suggests that interventions to promote more symmetric loading between limbs during these tasks may be broadly applicable for LLPUs.

During all tasks, the prosthetic ankle of LLPUs provided a limited amount of ankle RoM (Figs. 5-6). Sit-to-stand, squatting, and lifting are all tasks known to involve a high degree of biological ankle flexion (Grimmer et al., 2020). Our control participants had 19° of ankle RoM during sit-to-stand and >30° during squatting and lifting (Figs. S5 and S6, Table S2). In contrast, the prosthetic ankle provided <10° of ankle RoM for all tasks (Figs. 5-6, Table 2). Interestingly, the intact (biological) ankle of prosthesis users seems to also have reduced RoM compared to the ankles of controls. Even so, the intact ankle of LLPUs had more than twice the RoM of their prosthetic ankle. The large difference in RoM between the intact and prosthetic ankle might be a contributing factor to why prosthesis users overload their intact limb. Inadequate dorsiflexion may hinder users from comfortably positioning their prosthetic limb and trusting its ability to handle loading during these tasks. A prosthetic ankle-foot with increased dorsiflexion capabilities might allow LLPUs to position their lower limbs more symmetrically and put more force through their prosthetic limb while standing up or squatting down, but this requires further investigation.

Future work should investigate interventions that promote symmetric loading during sit-to-stand, squatting, and lifting to increase the safety and ability of LLPUs to complete these tasks. Interventions that involve lower limb muscle strengthening or task-specific training should be considered in addition to prosthetic device design approaches. Future work should also investigate lower limb loading during daily tasks in a broader population of LLPUs. One limitation of this study is all participants are highly active, unilateral transtibial LLPUs. Less active prosthesis users who find sit-to-stand, squatting, and lifting challenging or even impossible may exhibit different loading asymmetries and overall movement patterns, although we expect they would still accomplish these tasks by overloading their intact limb. Additionally, we focused on a limited set of biomechanics metrics. We characterized limb loading using peak vertical GRFs and knee flexion moments. We did not characterize frontal plane knee moments, which are often used as an indicator of knee loading and osteoarthritis risk during walking. During bilateral tasks (like the tasks presented in this study), frontal plane knee moments are relatively small due to a less dynamic shift in the center of pressure from side to side (Nagura et al., 2002; Trepczynski et al., 2014). Instead, increases in the sagittal plane knee moment have been shown to correlate to increased knee contact forces during these tasks (Ding et al., 2023; Trepczynski et al., 2014). While we assessed lower limb loading and ankle RoM, LLPUs may exhibit other compensation strategies during these tasks, such as altered trunk mechanics. Future research should build on this work to further characterize LLPU biomechanics during daily tasks.

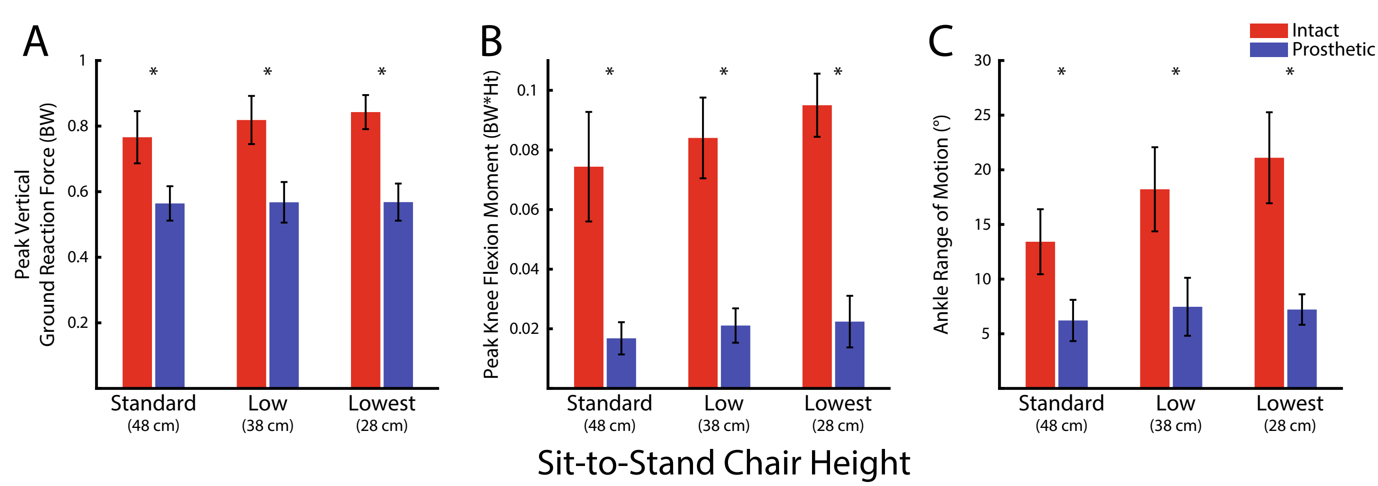

Fig. 6. (A) Peak vertical ground reaction force, (B) peak knee flexion moment, and (C) ankle range of motion during sit-to-stand from three chair heights: Standard (48 cm), Low (38 cm), and Lowest (28 cm). Intact limb values are in red and prosthetic limb values are in blue. Participant 7 could not complete the Lowest condition, so the data presented for that task is an average from $N = 7$. The Standard and Low tasks are averages from $N = 8$. Force was normalized to body weight (BW) and moment was normalized to body weight and height (Ht). *A significant difference was detected between the intact limb (red) and the prosthetic limb (blue) for all tasks. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

5. Conclusions

Unilateral transtibial LLPUs asymmetrically load their lower limbs during sit-to-stand, squatting, and lifting by overloading their intact limb compared to their prosthetic limb. The increased load on the intact limb was evident through significantly greater peak vertical GRFs and knee flexion moments. This asymmetric loading could be a contributing factor to the increased risk of musculoskeletal injury, joint degeneration, and pain in the intact limb of LLPUs. During all tasks, the prosthetic ankle provided significantly less ankle RoM than the intact ankle. Future work should investigate rehabilitation and prosthetic device interventions that promote symmetric loading to determine how this impacts user performance during these tasks.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.clinbiomech.2023.106041.

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