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Can static optimization detect changes in peak medial knee contact forces induced by gait modifications?

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ABSTRACT

Medial knee contact force (MCF) is related to the pathomechanics of medial knee osteoarthritis. However, MCF cannot be directly measured in the native knee, making it difficult for therapeutic gait modifications to target this metric. Static optimization, a musculoskeletal simulation technique, can estimate MCF, but there has been little work validating its ability to detect changes in MCF induced by gait modifications. In this study, we quantified the error in MCF estimates from static optimization compared to measurements from instrumented knee replacements during normal walking and seven different gait modifications. We then identified minimum magnitudes of simulated MCF changes for which static optimization correctly identified the direction of change (i.e., whether MCF increased or decreased) at least 70% of the time. A full-body musculoskeletal model with a multi-compartment knee and static optimization was used to estimate MCF. Simulations were evaluated using experimental data from three subjects with instrumented knee replacements who walked with various gait modifications for a total of 115 steps. Static optimization underpredicted the first peak (mean absolute error = 0.16 bodyweights) and overpredicted the second peak (mean absolute error = 0.31 bodyweights) of MCF. Average root mean square error in MCF over stance phase was 0.32 bodyweights. Static optimization detected the direction of change with at least 70% accuracy for early-stance reductions, late-stance reductions, and early-stance increases in peak MCF of at least 0.10 bodyweights. These results suggest that a static optimization approach accurately detects the direction of change in early-stance medial knee loading, potentially making it a valuable tool for evaluating the biomechanical efficacy of gait modifications for knee osteoarthritis.

1. Introduction

Knee osteoarthritis is a leading cause of disability (Vos et al., 2012) and affects over 654 million individuals worldwide (Cui et al., 2020). The disease is often isolated to the medial compartment, which has 5–10 times higher prevalence compared to lateral knee osteoarthritis (Ahlbäck, 1968; Felson et al., 2002; Jones et al., 2013). Andriacchi et al. (2004) have suggested that medial knee osteoarthritis is associated with compressive medial loading. The knee adduction moment is an easier-to-estimate surrogate for compressive medial knee contact force (MCF), and relates to medial knee osteoarthritis progression (Miyazaki et al., 2002) and severity (Sharma et al., 1998). Therefore, gait modifications have aimed to reduce the peak knee adduction moment (Fregly et al., 2007; Shull et al., 2013; Simic et al., 2011). Recently, MCF has been related to cartilage loss, and thus medial knee osteoarthritis progression (Brisson et al., 2021). Since changes in the knee adduction moment do not fully describe changes in MCF (Meyer et al., 2013b; Richards et al., 2018; Walter et al., 2010), load-reducing interventions may be more effective if they aim to reduce MCF (Kinney et al., 2013b).

Musculoskeletal simulation is often used to estimate MCF. Although there are various methods to estimate joint loading (Lloyd and Besier, 2003; Pizzolato et al., 2015), static optimization is a simple and commonly used technique. This method solves for muscle forces by optimizing muscle activations based on ground reaction force and

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Table 1
Demographics and number of steps per walking pattern evaluated for three Grand Challenge subjects.

	3rd Grand Challenge	4th Grand Challenge	6th Grand Challenge
Demographics (Fregly et al., 2012)			
Leg alignment	Neutral	Neutral	Valgus
Leg with instrumented knee	Left	Right	Right
Body mass index (kg/m^2)	28.1	23.6	23.7
Gender	Female	Male	Male
Number of steps per walking pattern			
Baseline	5	5	4
Bouncy	4	6	6
Crouch	4	Mild (6)	5
Forefoot strike	5	6	4
Medial thrust	5	8	N.A.*
Smooth	5	N.A.*	2
Trunk sway	4	N.A.*	N.A.*
Walking poles	Long (2)	Long narrow (6)	N.A.*
	Short (3)	Long wide (6)	
		Short narrow (7)	
		Short wide (7)	

* Not Available. The given walking pattern was not included in the dataset for the given subject.

motion data. It is important to test the accuracy of simulation estimates of MCF against measurements from instrumented knee replacements (D'Lima et al., 2012; Hicks et al., 2015). In the context of gait modifications, it is critical to evaluate "directional accuracy," which we define as the accuracy of correctly predicting whether an increase or decrease in MCF occurs from a gait modification compared to natural walking. Previous studies have reported MCF and total knee contact force errors during stance phase using static optimization (Brandon et al., 2014; DeMers et al., 2014; Lerner et al., 2015; Lundberg et al., 2013), but there has been less work testing the ability of static optimization to detect changes in MCF across multiple individuals and gait modifications.

To address this gap in knowledge, we sought to: (1) quantify the errors in MCF estimated from static optimization and (2) identify when static optimization can accurately determine whether a gait modification increases or decreases MCF. As an exploratory aim, we also evaluated the accuracy with which static optimization estimates lateral contact force and total contact force to provide a more complete analysis of knee joint loading. All simulations used in this study are freely available at https://simtk.org/projects/statop_val.

2. Methods

2.1. Overview

We simulated knee contact force using a musculoskeletal model and a custom static optimization implementation for three individuals and seven gait modifications. To compare the contact forces estimated from static optimization (i.e., simulated forces) and experimental contact forces, we evaluated mean absolute error at peaks of contact force and root mean square error over stance phase. We also determined the accuracy of correctly predicting the direction of change in peak contact force induced by gait modifications (i.e., whether force increased or decreased), which we call directional accuracy.

2.2. Experimental data

From the six Grand Challenge competitions for predicting knee loads (Fregly et al., 2012; Kinney et al., 2013a), we selected three subjects who performed the greatest variety of gait modifications. We analyzed data from the 3rd, 4th, and 6th competitions, including instrumented knee forces, motion capture, and force plate data from overground gait trials for natural walking (baseline) and seven different gait modifications (Table 1).

Instrumented knee replacements measured total contact force, and we calculated MCF and lateral contact force using previously published regression equations (3rd Grand Challenge: Fregly et al., 2012; 4th Grand Challenge: Zhao et al., 2007; 6th Grand Challenge: Meyer et al., 2013a). Simulated and measured knee contact forces were low-pass filtered at 15 Hz (zero-lag, 4th order Butterworth). We removed trials with erroneous experimental marker, ground reaction force, and instrumented knee data (nine of 124 trials).

The experimental data were collected using three sequential force plates. For some trials, the foot of the limb with the instrumented implant did not contact the middle plate. For these trials, we simulated a full stance phase by combining simulation results from the first 50% of stance at the last plate with the last 50% of stance at the first plate (Rajagopal et al., 2016). We defined the first and second contact force peaks as the maximum value between 15–35% and 65–85% of stance, respectively. We selected these ranges as $\pm$ 10% around 25% and 75% of stance, which are commonly used for defining early-stance and late-stance peaks (Fregly et al., 2009).

2.3. Musculoskeletal simulation

We integrated a multi-compartment knee (Lerner et al., 2015) into a full-body model (Rajagopal et al., 2016) to compute medial and lateral compartment contact forces. Using the Scaling Tool in OpenSim 4.0 (Seth et al., 2018), we used standing static calibration poses to scale each subject's model. We evaluated bilateral frontal-plane knee alignment and tibiofemoral contact points on the implanted limb from a standing radiograph for the 6th Grand Challenge subject (Lerner et al., 2015). Radiographs were not available for the other subjects, so we determined contact points using regression equations with femoral epicondyle marker positions as inputs (Winby et al., 2009). We used the OpenSim Inverse Kinematics Tool to solve for the kinematics that minimized the sum-squared error between experimental and virtual markers.

We used a custom static optimization implementation in MATLAB R2017b (Mathworks Inc., Natick, MA, USA) utilizing OpenSim; this implementation includes passive muscle forces and tendon compliance (Uhlrich et al., 2022). Using MATLAB's *fmincon* function, a constrained nonlinear solver, the objective function minimized the sum of squared muscle activations (Anderson and Pandy, 2001). We used the OpenSim Joint Reaction Analysis Tool to estimate MCF and lateral contact force from static optimization muscle force estimates; total contact force was the sum of these compartmental loads.

2.4. Error evaluation

We compared simulated and measured knee contact forces to evaluate the simulation results. We calculated mean absolute error at the two peaks of medial, lateral, and total contact force during the stance phase for all trials of baseline and gait modifications ($N = 115$ trials), and also for changes from baseline induced by gait modifications ($N = 101$ trials). We calculated root mean square error of medial, lateral, and total contact force over stance phase and averaged over all trials.

It is important to identify the magnitudes of change in contact force that static optimization can accurately detect. To identify these magnitudes, we first calculated simulated changes in contact force peaks by subtracting the subject's average simulated baseline peak (across all trials of normal walking) from the simulated peak in each gait modification trial, in both early and late stance. We used the same procedure for calculating measured changes in peaks from experimental forces. Thus, positive changes represent increases from baseline, and negative changes represent decreases from baseline.

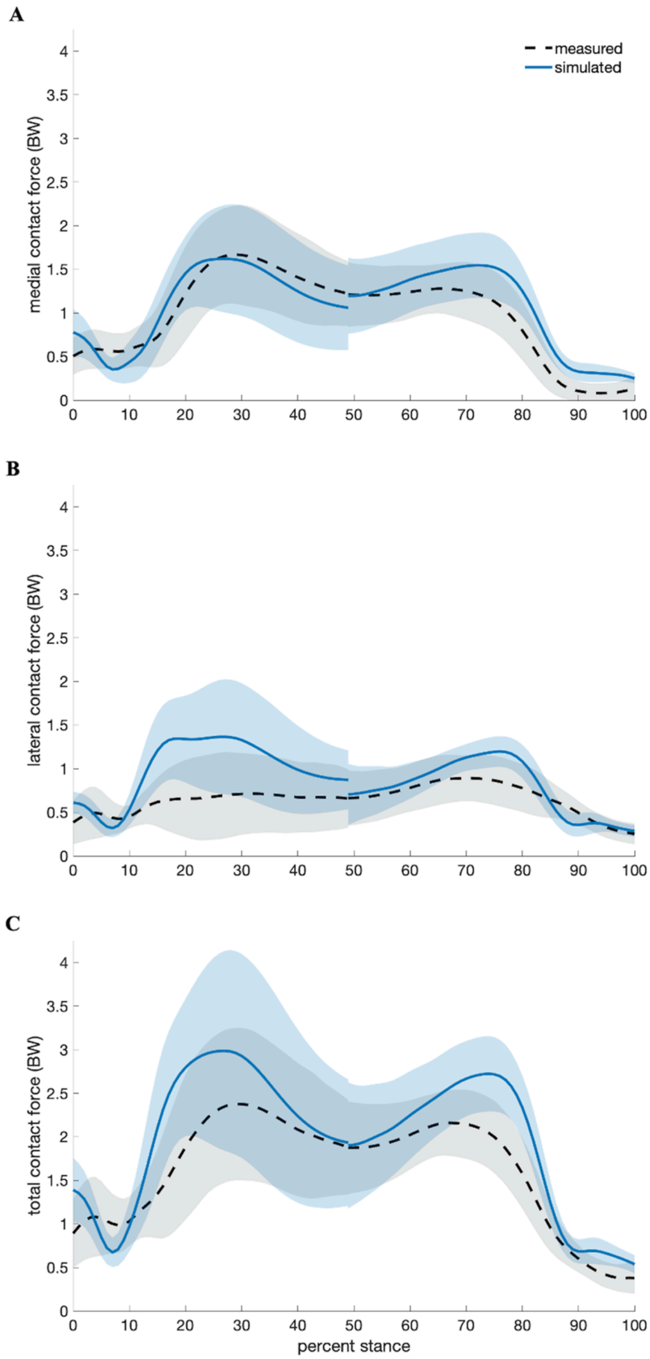

Fig. 1. Mean ± standard deviation (shaded) of measured and simulated medial (**A**), lateral (**B**), and total (**C**) knee contact force in bodyweights (BW) for all evaluated trials ($N = 115$). The discontinuity at 50% of stance results from computing early-stance and late-stance contact forces from different gait cycles for some steps.

We evaluated 0.05 bodyweight (BW) increments of simulated change from 0 to ± 1 BW. At each increment that contained more than three steps, we evaluated all steps for which static optimization predicted a change greater than the increment (i.e., simulated change thresholds). The "directional accuracy" was the proportion of these steps for which static optimization predicted the same direction of change as the experimental data (i.e., both positive or both negative). We also calculated overall directional accuracies at each contact force peak for all steps.

To establish a target directional accuracy for the simulated contact forces, we defined the directional accuracy of the standard practice for gait modifications, which is to prescribe the same load-reducing modification to all participants. This non-personalized approach assumes that a given modification has the same peak-changing effect on all individuals (Hunt et al., 2018; Simic et al., 2011). Based on previous literature (Fregly et al., 2009; Kinney et al., 2013b; Razu and Guess, 2018; Simic et al., 2012; Steele et al., 2012), we assumed that all gait modifications used in our study should decrease both peaks of medial, lateral, and total knee contact force, except for the first peak of bouncy gait and both peaks of crouch gait, which were assumed to increase. For each gait modification trial in the instrumented knee data, we calculated the peak change by subtracting the average baseline peak from the gait modification trial peak. We calculated directional accuracy as the proportion of peak changes that matched the assumed direction of change from the literature. A naïve classifier using these assumptions would be at most 68% accurate (Supplementary Table 1). Therefore, we defined *sufficient* directional accuracy as 70% and identified the minimum simulated change threshold that achieved this accuracy for increases and decreases in both peaks of contact force estimates.

Table 2
Mean absolute errors (in bodyweights, BW) for both peaks of medial, lateral, and total knee contact force (MCF, LCF, TCF) per walking pattern, listed in order of decreasing sample size (n).

Walking pattern (n)	MCF Peak 1 (BW)	MCF Peak 2 (BW)	LCF Peak 1 (BW)	LCF Peak 2 (BW)	TCF Peak 1 (BW)	TCF Peak 2 (BW)
Walking poles (31)	0.09	0.19	0.92	0.41	0.88	0.56
Bouncy (16)	0.15	0.39	1.10	0.42	1.22	0.78
Crouch (15)	0.21	0.28	1.09	0.34	1.24	0.67
Forefoot strike (15)	0.24	0.38	0.75	0.41	0.84	0.79
Medial thrust (13)	0.13	0.26	0.88	0.32	0.77	0.51
Smooth (7)	0.29	0.29	0.18	0.37	0.42	0.60
Trunk sway (4)	0.20	0.42	0.37	0.28	0.16	0.73
Baseline (14)	0.13	0.47	0.33	0.44	0.31	0.71
All patterns (115)	**0.16**	**0.31**	**0.81**	**0.39**	**0.83**	**0.65**

3. Results

Static optimization underpredicted first peak MCF with mean absolute error of 0.16 BW, and overpredicted the second peak with mean absolute error of 0.31 BW (Fig. 1A, Table 2). The mean absolute error of changes in MCF from baseline was 0.18 BW at the first peak and 0.35 BW at the second peak. Average stance-phase root mean square error was 0.32 BW for MCF. Static optimization detected the direction of change with sufficient accuracy (≥70%) for 0.05 BW increases at the first peak and 0.05 BW decreases at the second peak of MCF (Fig. 2A, Supplementary Table 2). Sufficient accuracy was also achieved for 0.10 BW decreases at the first peak. However, sufficient accuracy was not achieved for second peak increases.

Static optimization overpredicted both peaks of lateral and total contact force, with a first peak mean absolute error of 0.81 BW and 0.83 BW respectively, as well as 0.39 BW and 0.65 BW at the second peak (Fig. 1B & 1C). The larger first peak errors were primarily due to errors from the gait modification trials rather than the baseline trials (Table 2). The mean absolute errors of changes from baseline for lateral and total contact force at the first peak were 0.45 BW and 0.60 BW respectively, and 0.21 BW and 0.37 BW at the second peak. Average stance-phase root mean square error was 0.47 BW for lateral contact force and 0.66 BW for total contact force. The minimum simulated change thresholds that achieved sufficient directional accuracy for lateral and total contact force were larger than those for MCF, ranging from 0.15–0.45 BW at the first peak and 0.05–0.15 BW at the second peak (Fig. 2B & 2C,

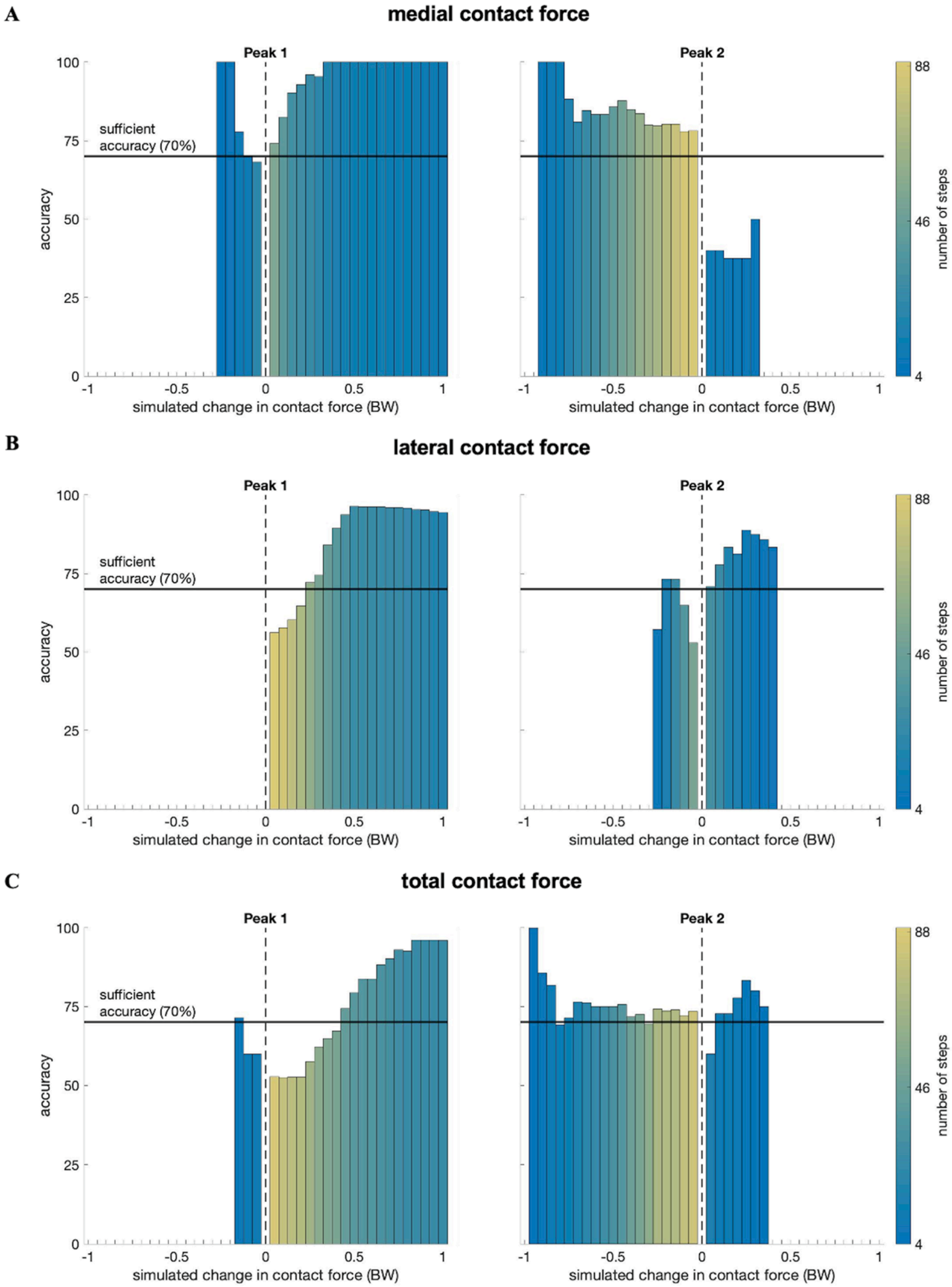

Fig. 2. Directional accuracy (how accurately static optimization classifies an increase or decrease in knee contact force induced by gait modifications) versus simulated change at the first and second peaks of medial (**A**), lateral (**B**), and total (**C**) knee contact force in bodyweights (BW). Positive changes represent increases from baseline. The color of each bar denotes the number of steps at or above each simulated change threshold. For example, for all gait modification steps that static optimization predicted to be at least a 0.05 BW reduction at Peak 1 of medial contact force (**A**, left), about 68% of them were actually reductions, according to the instrumented knee data. Sufficient directional accuracy is 70% (horizontal black line). The smallest simulated changes in knee contact force that achieved sufficient directional accuracy are reported in Supplementary Table 2.

Table 3
Proportions of experimental peak increases that were correctly predicted for medial, lateral, and total knee contact force (MCF, LCF, TCF) per gait modification, listed in order of decreasing total sample size (*n*).

Gait modification (*n*)	MCF Peak 1 (BW)	MCF Peak 2 (BW)	LCF Peak 1 (BW)	LCF Peak 2 (BW)	TCF Peak 1 (BW)	TCF Peak 2 (BW)
Walking poles (31)	8 / 10 (80%)	0 / 1 (0%)	12 / 12 (100%)	0 / 7 (0%)	6 / 6 (100%)	0 / 3 (0%)
Bouncy (16)	14 / 14 (100%)	1 / 4 (25%)	16 / 16 (100%)	6 / 7 (85.7%)	14 / 14 (100%)	1 / 3 (33.3%)
Crouch (15)	10 / 12 (83.3%)	2 / 4 (50%)	8 / 8 (100%)	7 / 9 (77.8%)	11 / 11 (100%)	3 / 5 (60%)
Forefoot strike (15)	3 / 8 (37.5%)	1 / 6 (16.7%)	9 / 13 (69.2%)	2 / 9 (22.2%)	6 / 9 (66.7%)	3 / 7 (42.9%)
Medial thrust (13)	6 / 7 (85.7%)	0 / 7 (0%)	5 / 5 (100%)	4 / 7 (57.1%)	6 / 7 (85.7%)	1 / 7 (14.3%)
Smooth (7)	2 / 3 (66.7%)	1 / 3 (33.3%)	2 / 2 (100%)	2 / 7 (28.6%)	3 / 3 (100%)	0 / 4 (0%)
Trunk sway (4)	1 / 4 (25%)	N.A.*	1 / 1 (100%)	3 / 4 (75%)	2 / 3 (66.7%)	1 / 2 (50%)
All modifications (101)	**44 / 58 (75.9%)**	**5 / 25 (20%)**	**53 / 57 (93%)**	**24 / 50 (48%)**	**48 / 53 (90.6%)**	**9 / 31 (29%)**

* Not Applicable. Zero experimental changes at the given knee contact force and peak type.

Table 4
Proportions of experimental peak reductions that were correctly predicted for medial, lateral, and total knee contact force (MCF, LCF, TCF) per gait modification, listed in order of decreasing total sample size (*n*).

Gait modification (*n*)	MCF Peak 1 (BW)	MCF Peak 2 (BW)	LCF Peak 1 (BW)	LCF Peak 2 (BW)	TCF Peak 1 (BW)	TCF Peak 2 (BW)
Walking poles (31)	10 / 21 (47.6%)	30 / 30 (100%)	0 / 19 (0%)	15 / 24 (62.5%)	2 / 25 (8%)	28 / 28 (100%)
Bouncy (16)	0 / 2 (0%)	12 / 12 (100%)	N.A.*	6 / 9 (66.7%)	0 / 2 (0%)	11 / 13 (84.6%)
Crouch (15)	2 / 3 (66.7%)	11 / 11 (100%)	0 / 7 (0%)	2 / 6 (33.3%)	0 / 4 (0%)	10 / 10 (100%)
Forefoot strike (15)	5 / 7 (71.4%)	5 / 9 (55.6%)	0 / 2 (0%)	6 / 6 (100%)	3 / 6 (50%)	6 / 8 (75%)
Medial thrust (13)	3 / 6 (50%)	5 / 6 (83.3%)	4 / 8 (50%)	4 / 6 (66.7%)	1 / 6 (16.7%)	6 / 6 (100%)
Smooth (7)	0 / 4 (0%)	3 / 4 (75%)	1 / 5 (20%)	N.A.*	1 / 4 (25%)	1 / 3 (33.3%)
Trunk sway (4)	N.A.*	4 / 4 (100%)	0 / 3 (0%)	N.A.*	0 / 1 (0%)	2 / 2 (100%)
All modifications (101)	**20 / 43 (46.5%)**	**70 / 76 (92.1%)**	**5 / 44 (11.4%)**	**33 / 51 (64.7%)**	**7 / 48 (14.6%)**	**64 / 70 (91.4%)**

* Not Applicable. Zero experimental changes at the given knee contact force and peak type.

Supplementary Table 2). Overall, static optimization predicted medial, lateral, and total contact force with higher directional accuracy for increases at the first peak and reductions at the second peak (Tables 3 & 4).

4. Discussion

Our primary objective was to determine how accurately static optimization estimates changes in peak knee MCF induced by gait modifications, and secondarily, to explore the accuracy of static optimization for estimating peak lateral contact force and peak total contact force. We found that if static optimization detected early-stance reductions, late-stance reductions, and early-stance increases in MCF of at least 0.10 BW, it was more accurate than the naïve assumption that gait modifications affect all individuals the same way. If static optimization detects changes greater than this threshold, we can be reasonably confident that the direction of the change is correct. This finding suggests that static optimization may be useful for evaluating how gait modifications affect early-stance MCF in future studies of individuals with osteoarthritis.

Our average root mean square errors are within the ranges of past Grand Challenge winners for best predictions (MCF: 0.21–0.47 BW; lateral contact force: 0.26–0.51 BW; total contact force: 0.48–0.69 BW), who used a variety of modeling and simulation techniques, including those that incorporated deformable contact models or electromyography signals (Hast and Piazza, 2013; Jung et al., 2016; Kim et al., 2013; Kinney et al., 2013a; Knowlton et al., 2013; Manal and Buchanan, 2013). This suggests that a simple muscle redundancy solver has comparable error to more complex approaches for estimating knee contact force. Additionally, although peak total contact force error was greater than peak MCF error, it is similar to those reported previously using comparable modeling and simulation methods. Our peak total contact force errors (first peak: 0.83 BW, second peak: 0.65 BW) and average stance-phase root mean square error (0.66 BW) are within ranges reported by Knarr and Higginson (first peak: 0.14–0.95 BW; second peak: 0.40–1.37 BW; root mean square error: 0.37–0.67 BW), who used a default static optimization implementation in OpenSim and simulated for the first three Grand Challenge subjects (2015).

Our static optimization implementation has sufficient sensitivity for identifying clinically relevant reductions in peak total contact force and MCF derived from weight loss studies. A 7.7–10.2% BW loss has been shown to improve function and reduce knee joint loading (Atukorala et al., 2016; Messier et al., 2011). Additionally, peak total contact force reduces 1–4-fold with every unit of weight loss (Aaboe et al., 2011; DeVita et al., 2016; Messier et al., 2005). Applying this ratio to 7.7–10.2% BW loss, we expect peak total contact force reductions in the range of 0.08–0.41 BW to be clinically meaningful. Our simulations detected reductions in peak total contact force at the lower end of this range (0.05–0.15 BW) with at least 70% accuracy (Supplementary Table 2). Similarly, since peak MCF comprised about 60–70% of peak total contact force in the Grand Challenge instrumented knee data, we expect that 0.05–0.29 BW peak MCF reductions are clinically meaningful. Our simulations detected peak MCF reductions at the lower end of this range (0.05–0.10 BW) with at least 70% accuracy.

It is important to identify the limitations of our study. Although a static optimization cost function that minimized sum-squared muscle activation resulted in sufficient directional accuracy for most cases (Fig. 2, Supplementary Table 2), this cost function may not optimally represent muscle coordination for individuals with osteoarthritis learning a new gait pattern (Shull et al., 2015). Incorporating additional cost function terms that better capture elevated co-contraction (Booij et al., 2020) or pain minimization, or using electromyography data to inform simulations (Lloyd and Besier, 2003; Pizzolato et al., 2015), might increase the accuracy of static optimization. Additionally, static optimization was not able to detect increases in second peak MCF with greater accuracy than a naïve approach. This may be due to the small sample size (25 steps) for second peak MCF increases (Table 3), and the greater number of possible muscle coordination patterns in late stance

(DeMers et al., 2014). Finally, since we calculated experimental MCF and lateral contact force from regression equations, they are not direct measurements. Our errors may be due, in part, to this indirect comparison.

This study demonstrates that a simple simulation pipeline can detect reductions and early-stance increases in peak knee MCF from gait modifications with at least 70% accuracy. When coupled with mobile sensing techniques for estimating kinematics and kinetics (Karatsidis et al., 2016), static optimization might be used to evaluate joint loading outside of the lab. The accessibility of these tools and our open-source simulation software (Delp et al., 2007) can enable estimates of joint contact force to be used for evaluating gait modifications in future studies and in clinical practice.

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Data availability statement

Simulation results and input data needed to replicate the simulation results in this study are available at https://simtk.org/projects/statop_val.

CRediT authorship contribution statement

Janelle M. Kaneda: Writing – review & editing, Writing – original draft, Software, Methodology, Formal analysis, Conceptualization. **Kirsten A. Seagers:** Writing – review & editing, Software, Methodology, Conceptualization. **Scott D. Uhlrich:** Supervision, Conceptualization, Methodology, Software, Writing – review & editing. **Julie A. Kolesar:** Writing – review & editing, Software, Formal analysis. **Kevin A. Thomas:** Writing – review & editing, Methodology. **Scott L. Delp:** Writing – review & editing, Supervision, Resources, Conceptualization.

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.

Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.jbiomech.2023.111569.

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