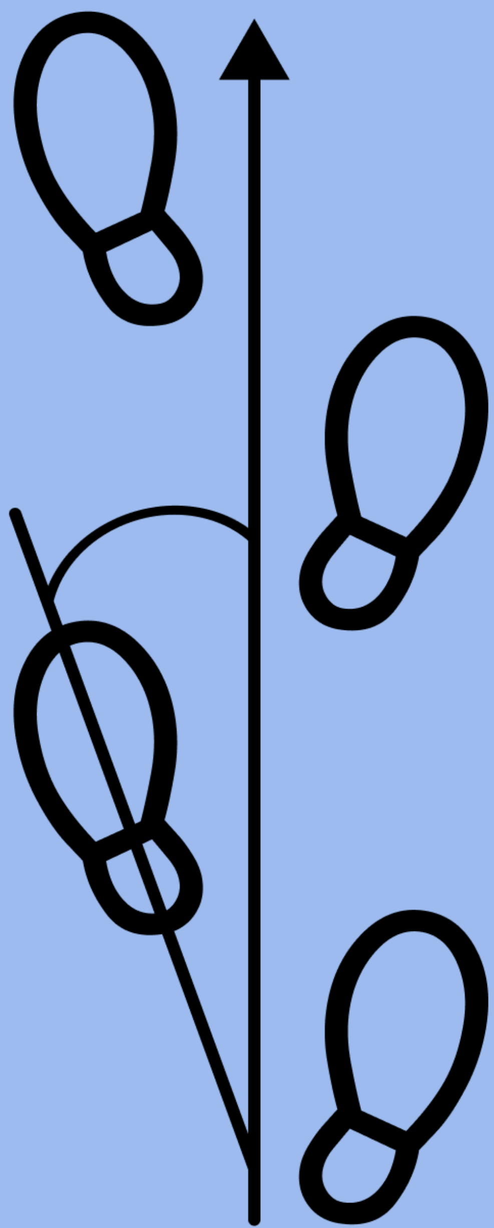
Altering the Foot Progression Angle During Gait: Effects in Kinematics and Estimated Knee Joint Loading

Thesis report

Line Antoinette Torsij



TU Delft

Altering the Foot Progression Angle During Gait: Effects in Kinematics and Estimated Knee Joint Loading

Thesis report

by

Line Antoinette Torsij

in partial fulfilment of the requirements of Master of Science in Biomedical Engineering, Track Neuromusculoskeletal Biomechanics, at the Delft University of Technology, to be defended publicly on Thursday December 18, 2025 at 10:30.

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TU Delft

Preface

This thesis marks the end of my time as a student, first during my bachelor's degree in Clinical Technology, and later during my master's degree in Biomedical Engineering. It has been quite a journey, with a lot of challenges, learning moments, and great memories along the way.

I would like to thank my daily supervisors, Mariska Wesseling and Erin Macri, for always making time to discuss any questions or problems I had. Your help, advice, and feedback made a big difference throughout this project. I also want to thank Jaap Harlaar for sharing his knowledge and giving valuable guidance. A big thank you as well to everyone from the Clinical Biomechanics group for your input and ideas during the lab meetings.

And finally, I want to thank all the friends and fellow students who have been part of my study years. You have made the past six years unforgettable, and I am really grateful that I got to share this time with you.

Line Antoinette Torsij
Delft, December 2025

Abstract

Background

Altering the foot progression angle (FPA) is a commonly researched approach to reduce the knee adduction moment (KAM) in patients with medial compartment knee osteoarthritis. However, little is known about kinematic strategies used to achieve specific FPAs, and the effect of different FPAs on the medial tibiofemoral contact force (TFCF).

Objective

This study aimed to investigate how walking with different FPAs affects lower-limb joint angles and medial TFCF, to assess whether a personalized target FPA can reduce medial TFCF, and to investigate the relationship between changes in KAM and changes in medial TFCF.

Methods

Motion capture data were recorded for healthy participants while walking with their natural FPA, as well as walking with four altered FPA conditions (toe-out and toe-in 5° and 10°, relative to the natural FPA). For each condition, the KAM was calculated using ground reaction forces and motion capture data, and joint angles and medial TFCF were obtained by performing musculoskeletal modeling. Peak values during the first and second part of the stance phase, as well as the area under the curve during stance phase (impulse), were extracted for the KAM and medial TFCF.

Results

Fourteen healthy participants (9 women, age: 57.8 ± 12.1 years) were included. Across all joint angles, changes for toe-out conditions occurred in opposite directions from toe-in conditions, with larger differences for 10° than 5° conditions. Hip rotation differed significantly throughout the stance phase, subtalar inversion/eversion showed significant differences during late stance. Changes in medial TFCF varied across FPA conditions without a consistent pattern. The personalized target FPA significantly reduced first peak (-0.22 ± 0.19 body weight (BW), $p = 0.002$), second peak (-0.23 ± 0.32 BW, $p = 0.049$), and impulse (-0.15 ± 0.13 BW, $p = 0.002$) medial TFCF. Changes in KAM were more strongly associated with changes in medial TFCF for the first peak during toe-out gait ($R^2 = 0.64$ and 0.82 for 5° and 10°, respectively) and for the second peak during toe-in gait ($R^2 = 0.44$ and 0.76 for 5° and 10°, respectively).

Conclusions

Lower-limb joint angles were consistently altered for different FPA conditions, with the largest changes observed in hip rotation and subtalar inversion/eversion. The change in medial TFCF for FPA alterations was highly individual, with no consistent pattern across participants. A personalized target FPA significantly reduced medial TFCF throughout the stance phase. The relationship between changes in KAM and medial TFCF was dependent on stance phase, consistent with prior research, and also varied across FPA conditions. Studies including patients with knee osteoarthritis are needed to determine how FPA alterations influence medial compartment loading in the target population.

Contents

1 Introduction

1.1. Knee Osteoarthritis and Gait Modifications

Knee osteoarthritis (OA) is the most common type of arthritis, affecting around 365 million people worldwide [1]. It is a progressive degenerative disease of the knee joint, characterized by damage to the articular cartilage. This damage can result from age-related degeneration, or from excessive joint loading due to lifestyle factors such as physically demanding work [2]. Treatment ranges from non-surgical interventions, such as lifestyle modifications and pharmacological therapy, to surgical interventions like total knee replacement. With an aging population, the prevalence and societal burden of this disease are expected to increase [3]. This highlights the importance of research focusing on prevention of disease progression, as that could delay or even prevent the need for surgery [4].

The medial compartment of the knee joint is most commonly affected in knee OA [5]. Research has shown that reducing internal loading of the medial compartment can slow disease progression [6], and gait modifications aiming to reduce this medial compartment loading are being researched extensively [7, 8]. Gait modifications can be described as passive, such as knee braces, insoles and modified shoes [9, 10, 11], or as active gait modifications. Examples of such strategies are increasing trunk sway and step width [12, 13, 14]. Another commonly researched active gait modification is altering the foot progression angle (FPA). The FPA is defined as the angle between the walking direction, and the direction of the foot on the ground during stance phase (Figure 1.1). Several studies have researched this form of gait modification to reduce medial compartment loading. These include both single-session studies [15, 16, 17] and longer-term interventions where participants trained to walk with a personalized target FPA for a longer time-period [18, 19, 20, 6]. Evidence shows that the effect of a specific FPA alteration on medial compartment loading varies between individuals. For some, toe-in gait produces the largest reduction, whereas for others, a toe-out modification is more effective [6]. These findings underscore the need for a personalized approach when selecting a target FPA.

1.2. Modifying the FPA

Different combinations of lower-limb joint angles can result in the same FPA. Investigating these adaptations may help identify consistent movement patterns, which could ultimately support the development of improved FPA assessment methods. Currently, evaluating the effect of multiple FPAs on medial compartment loading requires separate experiments for each condition. This approach is time-consuming and limited to the measured angles, meaning that the true optimal FPA, which produces the greatest reduction in medial compartment loading, might fall between the tested FPAs. Some studies have proposed alternative approaches: for example, one study created a patient-specific cost function to predict the influence of the FPA on knee joint moments using, among other inputs, differences in joint angles [21]. Although this approach works well for individual participants, it is unclear whether joint angle adaptations follow consistent patterns across people. Identifying these generalizable adaptations is important for developing predictive methods that can be applied to more individuals, allowing the effects of different FPAs to be evaluated using only natural gait.

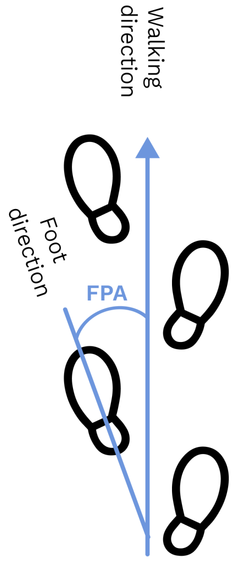


Figure 1.1. The foot progression angle (FPA). The FPA is calculated as the angle between the walking direction and direction of the foot on the ground.

1.3. FPA and Knee Loading

While the goal of gait modifications is to reduce medial compartment loading, this metric cannot be directly measured in vivo. Therefore, the external knee adduction moment (KAM) is commonly used as a surrogate measure. Although the KAM is associated with cartilage degeneration and disease progression [22, 6], several studies have shown that changes in KAM do not reliably reflect changes in actual medial compartment loading [23, 24]. The KAM reflects a moment around the knee, and is influenced by the magnitude and direction of the ground reaction force (GRF) and its medial-lateral position relative to the knee joint center [25]. Medial compartment loading is influenced by multiple factors, including muscle forces, which are only partially reflected in the KAM. One study found that non-knee-spanning muscles contributed more consistently to the KAM, as these muscles primarily affect the external moment around the knee, increasing compression in one compartment and unloading the other. In contrast, the contributions of knee-spanning muscles were not well reflected in the KAM, because these muscles primarily influence the total knee joint load by compressing or unloading both compartments [26]. This indicates that when the total load may change due to muscle forces, focusing only on the KAM does not capture the full effects on medial compartment loading [24].

The previously mentioned studies investigating the effects of different FPAs and personal target FPAs have primarily reported the KAM as the outcome measure. As a result, the impact of these gait modifications on medial compartment loading remains unclear. While some studies have assessed the effects on medial compartment loading for toe-in gait [27, 28], they focused only on one altered FPA. Consequently, it remains unknown how walking with a personalized target FPA, toe-in or toe-out, affects medial compartment loading.

Since direct measurement of knee joint loading is not possible, static optimization is often used to obtain estimations. Static optimization is a musculoskeletal modeling technique that solves for muscle forces by for example minimizing muscle activations, using motion capture (MoCap) and GRF data. Models incorporating separate medial and lateral compartments within the knee are able to calculate the medial tibiofemoral contact force (mTFCF), which provides an estimation of medial compartment loading [29]. Previous research has shown that this approach can accurately detect the direction of changes in medial compartment loading, as reflected by mTFCF. These findings were based on comparisons between simulated results and measurements from an instrumented knee during several gait modifications [30]. However, this method has not been validated for different FPAs. Although true validation requires comparison to data measured by an instrumented knee, several studies have investigated the relationship between the KAM and mTFCF. These studies showed that the maximal mTFCF during the first half of stance (first peak) is more strongly associated with the KAM than during the second half (second peak), in both healthy individuals and patients with knee OA [31, 32]. Another study investigated how changes in mTFCF relate to changes in KAM for a toe-in FPA [28]. They found a moderate association between changes in first peak KAM and first peak mTFCF, and a weak association for the second peak and total area under the curve (impulse). Because knee-spanning muscle forces contribute less to medial compartment loading during early stance, frontal-plane external loading plays a larger role, which explains why the first peak typically shows a stronger association between the KAM and mTFCF [29]. By investigating this relationship for different FPAs, insights can be gained into whether the change in estimated mTFCF shows patterns consistent with prior literature.

1.4. Research Objectives

The first aim of this study was to investigate, in adults without knee OA, how pelvic, hip, knee, and ankle-foot joint angles differ throughout the gait cycle between natural walking and walking with different FPAs. It was hypothesized that all analyzed lower-limb joint angles would exhibit consistent patterns of change for each FPA condition.

The second aim of this study was to investigate how walking with different FPAs affects medial compartment loading, and whether adopting a personalized target FPA, based on the first peak, leads to a reduction in medial compartment loading. It was hypothesized that individuals would generally respond better to either a toe-in or toe-out FPA for each peak metric, and that adopting a target FPA would significantly reduce first peak and impulse medial compartment loading, consistent with previously reported reductions in the KAM [33].

The third aim of this study was to investigate the relationship between changes in KAM and changes in mTFCF when comparing natural walking with walking with different FPAs. It was hypothesized that, across all FPA conditions, KAM changes would be more strongly associated with first peak mTFCF changes than with second peak [28].

2 Methods

2.1. Participants

This study was conducted as part of a parent study, researching a remote patient monitoring system designed to measure the FPA at home [34]. MoCap data gathered during the baseline measurements were used for the present study. More information on the parent study can be found in Appendix A. For the present study, eligibility criteria were 1) the ability to walk unaided for at least 15 to 30 minutes, 2) absence of activity related knee pain and morning stiffness, 3) absence of traumatic knee injury or surgery in the past year, 4) absence of any neurological, rheumatological or metabolic conditions affecting mobility, 5) absence of any musculoskeletal conditions affecting ankles, hips, pelvis or lumbar spine, 6) BMI < 35, 7) absence of cognitive impairments and 8) unable to communicate in Dutch or English. Ethical approval was granted by the Erasmus MC Medical Ethics Review Committee and all participants provided written informed consent.

2.2. Data Collection

During a single-session experiment, MoCap data were collected in the Erasmus MC / TU Delft Motion Biomechanics & Imaging Lab. Marker data were recorded at a frequency of 100 Hz. GRF data were recorded at 1000 Hz, and filtered using a 4^{th} order low pass Butterworth filter with a cut-off frequency of 10 Hz. Additionally, data were collected using the WalkWell Gait reTrainer system, which provided real-time FPA feedback based on an inertial measurement unit (IMU) placed on the right foot [34]. For this reason, only right-foot data were analyzed. The WalkWell system was used only to calculate the natural FPA and to provide feedback during the measurements in the lab, all other analyses were based entirely on MoCap data.

Reflective markers were placed on the skin of the participant according to a modified Conventional Gait Model 2.5 Marker set (Figure 2.1) [35]. The participant was given a few minutes of walking on the treadmill, both to get used to the feeling of walking on the treadmill, and to determine a comfortable walking speed. This self-selected walking speed was used for all following trials. Each participant then performed 5 trials: one while walking naturally, and four while walking with an altered FPA. Each trial was about 3 minutes. During the natural walking trial, the natural FPA was calculated using the WalkWell system, as the average over all steps of the trial [36]. Participants were instructed to modify their FPA for the remaining four trials, which consisted of toe-out angles of 5° and 10° and toe-in angles of 5° and 10°, relative to their natural FPA. To allow for natural variation and measurement error, the target window was set to the target FPA, plus and minus 2°. The FPA was calculated as the average over five steps using the WalkWell system. After every set of five steps, participants received visual and audio feedback indicating whether they needed to increase toe-in, increase toe-out, or if their FPA was within the target window.

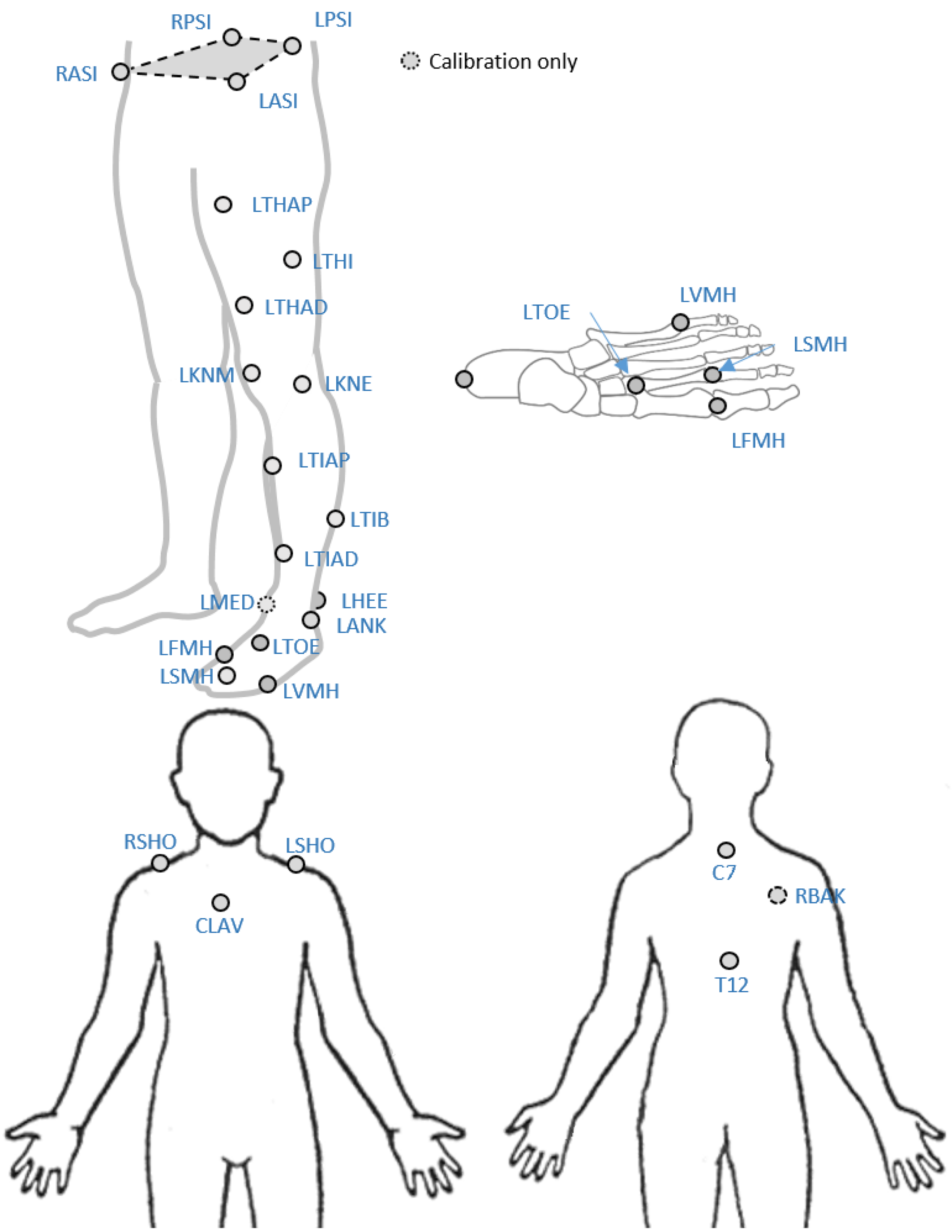


Figure 2.1. Modified Conventional Gait Model 2.5 Marker set [35]. Compared to the unmodified marker set, markers on the arms and head were omitted, the T2 marker was repositioned to C7, and the T10 marker was repositioned to T12.

2.3. Data Processing

GRF data were used to identify individual gait cycles, which were time-normalized to 100% gait cycle. For all five conditions, only the last 20 gait cycles were selected to ensure that participants had sufficient time to adapt to the FPA condition. Although real-time feedback on the FPA was provided in the lab using the WalkWell system, MoCap data were used to verify that only gait cycles within the target FPA window were included in further analysis. The direction of the foot on the treadmill during midstance was identified by comparing the location of the marker on the second metatarsal head (SMH) to the marker on the heel (HEE) (Figure 2.1). The FPA was then calculated as the difference between the treadmill's forward direction and the direction of the foot. If the FPA was outside the target window, that particular gait cycle was removed, resulting in fewer than 20 gait cycles for that specific condition.

2.4. Musculoskeletal Modeling

All valid gait cycles from each condition were analyzed using the open-source software OpenSim 4.5 [37, 38]. A model developed by Bedo [39] was used to estimate the mTFCF and lateral tibiofemoral contact force (lTFCF) during each gait cycle. This model integrated the knee model by Lerner [29] into a full body model. The Lerner knee model defines separate medial and lateral tibiofemoral compartments. Knee flexion/extension occurs via a single sagittal-plane hinge, which connects to the medial and lateral compartments via two parallel joints. These joints distribute all loads transmitted between the femur and tibia. This allows the model to resolve the contact forces in the medial and lateral compartments, the mTFCF and lTFCF, required to balance the muscle forces and internal and external loads acting on the model. Subject-specific frontal-plane orientation (i.e. knee varus-valgus) can be incorporated by specifying the alignment of the knee joint within the model. Subject-specific contact locations can be incorporated by shifting the medial and lateral compartment attachment points relative to the tibia. However, because radiographic imaging was not available for the present study, the frontal-plane alignment and compartment locations were kept at the neutral values defined in [29].

For each participant, the model was first scaled using a static pose, by iteratively checking the root mean square and maximal error of markers on anatomical landmarks. Joint angles were then calculated using the inverse kinematics tool, minimizing the sum of weighted squared errors of markers and coordinates [40]. Muscle activations and forces were calculated using static optimization, while minimizing the sum of squared muscle activations [41]. Coordinates were filtered with a cut-off frequency of 6 Hz. To allow the simulation to run properly, reserve actuators were appended to the model's force set. Lastly, a joint reaction analysis was performed to obtain the mTFCF and lTFCF [42].

2.5. Data Analysis

For each valid gait cycle, the KAM was calculated in the frontal plane as the cross product of the GRF, and the vector from the knee joint center (defined as the midpoint between the medial and lateral epicondyle markers) to the center of pressure on the treadmill belt. Results were then normalized by dividing the KAM by body weight. The following joint angles were extracted from inverse kinematics results: transverse pelvic rotation, hip abduction/adduction, hip rotation, hip flexion/extension, knee flexion/extension, ankle plantarflexion/dorsiflexion and subtalar inversion/eversion. The mTFCF and lTFCF were extracted from joint reaction analysis results by taking the vertical force component and dividing by body weight x 9.81 m/s^2. The total tibiofemoral contact force (tTFCF) was defined as sum of the mTFCF and lTFCF [39]. For each participant, joint angles, KAM, and contact forces were averaged across all valid gait cycles within each condition, resulting in mean curves for each variable.

The following parameters were extracted during stance phase for the mTFCF: first peak, second peak and impulse. Using these timings, the first peak and second peak KAM were extracted, as well as the first and second peak mTFCF expressed as a percentage of tTFCF (%tTFCF). The difference between natural walking and each FPA condition was calculated as ΔmTFCF, Δ%tTFCF and ΔKAM per participant per condition.

For each participant, a target FPA was selected as the condition resulting in the largest reduction in first peak mTFCF compared to natural walking [43]. If all conditions caused an increase, no target FPA was set.

2.6. Statistical Analysis

To address the first aim of this study, joint angles across the entire gait cycle were compared between each FPA condition and natural walking, using one-dimensional statistical parametric mapping (SPM). SPM produces a time series of t-values along the gait cycle, which allows identification of regions where joint angles differ significantly [44].

For the second aim, paired t-tests were performed to evaluate whether mTFCF and %tTFCF differed between the target FPA condition and natural walking. The analysis was done separately for the first peak, second peak, and impulse. Normality was checked using the Shapiro-Wilk test, and in case of violations, the non-parametric Wilcoxon signed-rank test was used.

For the third aim, a linear regression analysis was performed to assess the relation between the ΔKAM and ΔmTFCF at the same time point in the gait cycle, for each FPA condition separately. The fitted regression model, estimated using Ordinary Least Squares (OLS), was performed for the first peak, second peak and impulse values separately. The ΔKAM was treated as the independent variable, ΔmTFCF was treated as the dependent variable: $\Delta mTFCF = \beta_0 + \beta_1 \cdot \Delta KAM$. Normality, homoscedasticity and independence of residuals were checked both visually, and by using statistical tests; Shapiro-Wilk, Breusch-Pagan and Durbin-Watson, respectively.

Statistical significance was set to an alpha level of 0.05. All statistical analyses were conducted in Python (version 3.11.8) using the scipy.stats, statsmodels.formula.api and spm1d packages.

3

Results

3.1. Participant Demographics

A total of 14 participants (5 male, 9 female) were included in this study. The mean age was 57.8 ± 12.1 years, with an average height of 1.75 ± 0.11 m and weight of 72.8 ± 14.2 kg. The mean body mass index (BMI) was 23.7 ± 2.7 kg/m^2. The self-selected walking speed was 1.00 ± 0.19 m/s.

3.2. Alterations in Joint Angles

For all joint angles, a distinction between the toe-out and toe-in conditions was shown, where the 10° conditions generally produced larger differences than the 5° conditions (Figure 3.1). However, not all differences were statistically significant. The largest differences were observed in hip rotation, with significant differences present throughout the entire stance phase for all conditions. Subtalar inversion/eversion showed significant differences in both toe-out conditions and toe-in 10°, particularly during late stance. Differences in ankle plantarflexion/dorsiflexion were minimal at initial contact but increased during stance, with significant differences observed for the 10° FPA conditions. Transverse pelvic rotation remained nearly unchanged for toe-out conditions, whereas toe-in conditions exhibited a contralateral rotation in early stance, meaning the ipsilateral pelvis moved forward. Standard devations (SD) for transverse pelvic rotation were large for all FPA conditions. Hip abduction was increased during early stance for toe-out conditions. Hip and knee flexion differences were non-significant and less consistent throughout the stance phase.

3.3. Changes in Medial Compartment Loading

The pattern of ΔmTFCF varied across participants for all three parameters. For some participants, moderate increases for toe-out 5° were followed by larger decreases for toe-out 10°, whereas others exhibited inconsistent changes (Figure 3.2 and Appendix B). The direction and magnitude of change for a 5° alteration did not always match with the results for 10°. Additionally, both toe-in and toe-out conditions caused changes in the same direction for one participant, e.g. both increasing or decreasing. When comparing the first and second peaks, no distinct differences in overall response patterns were apparent. However, the spread of changes was greater for the first peak under toe-out conditions, whereas for the second peak, toe-in conditions showed larger differences. For the impulse, no such difference between conditions was observed.

Seven participants received a target FPA of toe-in 10°, one participant of toe-out 10° and two participants of toe-out 5°, as that condition resulted in the largest reduction in first peak mTFCF. The remaining four participants did not achieve a reduction with any of the FPA conditions (Appendix D). Because the normality assumption was violated for several groups, it was decided to perform the nonparametric Wilcoxon signed-rank test for all outcomes. All mTFCF parameters were significantly decreased using the target FPA (Table 3.1). For %tTFCF, significant reductions occurred for the first peak and impulse, while a non-significant increase was seen for the second peak.

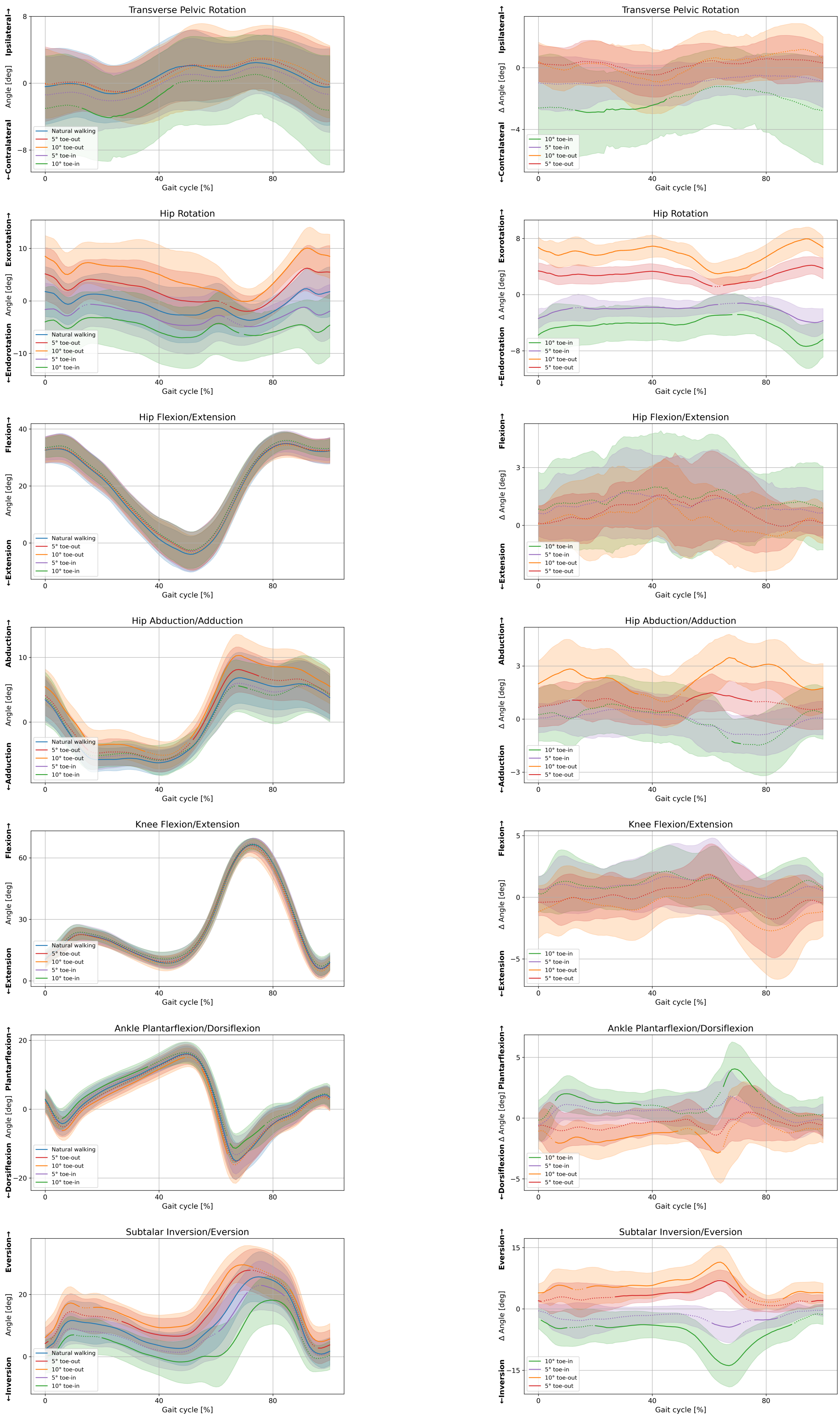


Figure 3.1. Joint angles for each FPA condition (left) and differences in joint angles between natural walking and each FPA condition (right) (mean[SD]). Solid lines indicate a significant difference between the FPA condition and natural walking (solid blue), dotted lines indicate non-significance.

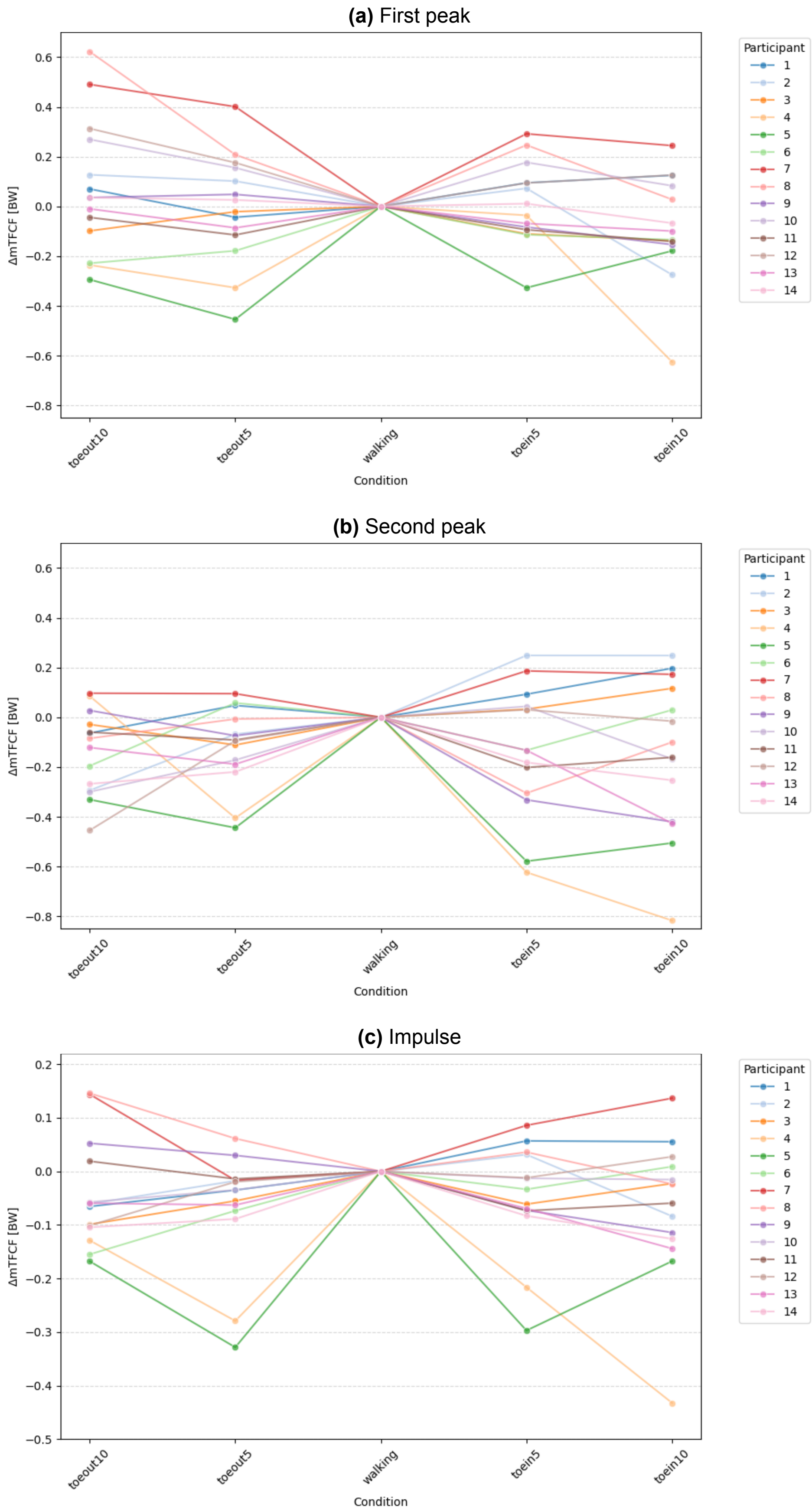


Figure 3.2. Difference in change in mTFCF (ΔmTFCF) between each FPA condition and natural walking, e.g. *walking*. Each line represents results for one participant. A negative change indicates a reduction compared to natural walking.

Table 3.1. Wilcoxon signed-rank test on differences between natural walking and the target FPA.

Outcome		Walking	Target	Difference	p
mTFCF [BW]	First Peak	2.09 ± 0.44	1.93 ± 0.32	-0.22 ± 0.19	**0.002**
	Second Peak	2.56 ± 0.53	2.40 ± 0.51	-0.23 ± 0.32	**0.049**
	Impulse	1.18 ± 0.31	1.07 ± 0.24	-0.15 ± 0.13	**0.002**
%tTFCF [%]	First Peak	79.6 ± 7.21	76.1 ± 7.51	-4.97 ± 5.93	**0.037**
	Second Peak	75.9 ± 6.26	76.2 ± 7.04	0.34 ± 4.91	0.695
	Impulse	72.1 ± 5.51	69.8 ± 5.45	-3.21 ± 3.97	**0.004**

Abbreviations. mTFCF = medial tibiofemoral contact force, %tTFCF = medial tibiofemoral contact force expressed as percentage of the total tibiofemoral contact force, BW = body weight.
Results are displayed as mean ± SD. *Walking* indicates mean results for natural walking over all participants. *Target* indicates mean results for the target FPA condition over all participants. *Difference* indicates the paired difference between the walking and target condition for all participants. Bold results indicate significance ($p < 0.05$).

3.4. Relationship Between ΔKAM and ΔmTFCF

The ΔKAM and ΔmTFCF were positively associated for all FPA conditions (Table 3.2, Figure 3.3). However, for first peak toe-in 10°, this association was not significant. The assumptions of normality or homoscedasticity were violated for second peak toe-in 10° and impulse toe-out and toe-in 5°. The estimated KAM throughout the stance phase for each participant and condition, as used in the linear regression analysis, is provided in Appendix C.

Table 3.2. Regression analysis results for the relationship between changes in KAM (ΔKAM) and changes in mTFCF (ΔmTFCF).

Measure	Condition		R^2	β_1 (BW per Nm/kg)
First peak	Toe-out	5°	0.64	**2.16**
		10°	0.82	**1.91**
	Toe-in	5°	0.36	**0.98**
		10°	0.29	1.20
Second peak	Toe-out	5°	0.29	**1.56**
		10°	0.34	**1.15**
	Toe-in	5°	0.44	**4.48**
		*10°	0.76	**4.65**
Impulse	Toe-out	* 5°	0.55	**2.58**
		10°	0.78	**1.78**
	Toe-in	* 5°	0.41	**1.77**
		10°	0.65	**2.64**

Bold results indicate significance ($p < 0.05$).
* Indicates that test assumptions were violated.

For the first and second peaks, the regression slopes, β_1 (BW per (Nm/kg)), were similar within each FPA direction (Table 3.2). For the first peak, toe-out conditions showed steeper slopes (β_1 of 2.16 and 1.91 for 5° and 10°, respectively) compared with toe-in conditions (β_1 of 0.98 and 1.20) (Figure 3.3a). For the second peak, this pattern was reversed, with steeper slopes for toe-in conditions (β_1 of 4.48 and 4.65) than for toe-out conditions (β_1 of 1.56 and 1.15) (Figure 3.3b). A difference in pattern for toe-in and toe-out conditions was not present for the impulse (Figure 3.3c).

Associations between ΔKAM and ΔmTFCF were stronger for the first peak in toe-out conditions (R^2 = 0.64 and 0.82 for 5° and 10°, respectively) compared with toe-in conditions (Table 3.2). For the second peak, toe-in conditions showed stronger associations (R^2 = 0.44 and 0.76 for 5° and 10°, respectively).

Across all parameters, associations were generally stronger for the 10° than the 5° conditions, except for the first peak in toe-in gait, where R^2 was 0.36 for 5° and 0.29 for 10°.

For the first peak, toe-in results were generally located to the left relative to toe-out results (Figure 3.3a). At comparable ΔmTFCF values, toe-in results corresponded to lower ΔKAM values than toe-out results. For the second peak, the opposite pattern was shown, with toe-out results located to the left of toe-in results (Figure 3.3b). For both the first and second peaks, points were more widely distributed at lower ΔKAM and ΔmTFCF values, whereas the distribution was more compact at higher values. Toe-in and toe-out results did not form distinct clusters for the impulse, and the results were distributed more evenly across the range of ΔKAM and ΔmTFCF values (Figure 3.3c).

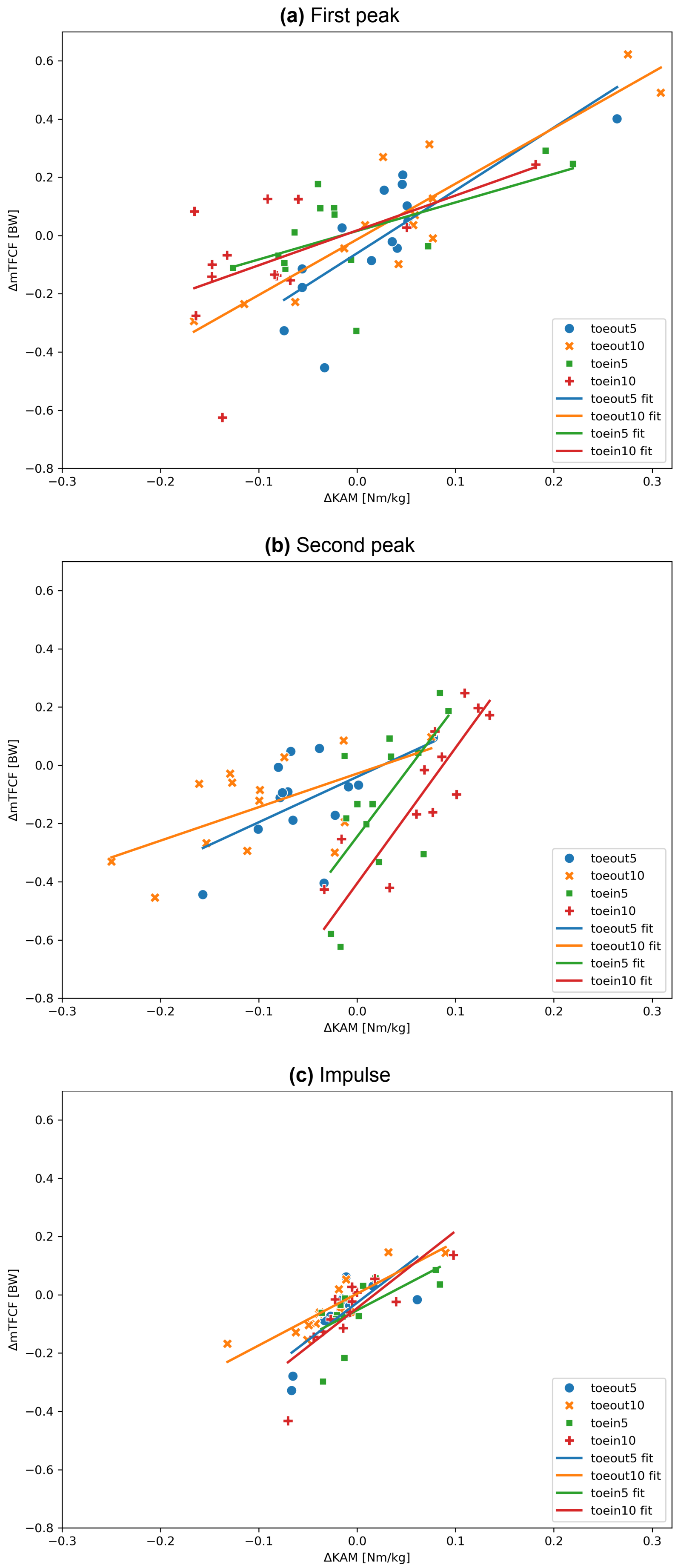


Figure 3.3. Scatterplot and fitted model of changes in KAM (ΔKAM) against changes in mTFCF (ΔmTFCF) for all participants and conditions. Colors and markers indicate FPA conditions.

4 Discussion

This study aimed to investigate 1) how lower-limb joint angles differ throughout the gait cycle between natural walking and different FPAs, 2) how walking with different FPAs alters medial compartment loading and whether adopting a personalized target FPA leads to a reduction in medial compartment loading, and 3) the relationship between changes in KAM and mTFCF when comparing natural walking with walking with different FPAs.

4.1. Alterations in Joint Angles

Hip rotation and subtalar inversion/eversion showed the clearest distinction between the natural walking, toe-in, and toe-out conditions. Together, their alterations accounted for nearly the full change in FPA, suggesting that most of the rotation required to modify the FPA originates from the hip and subtalar joint. Although differences in the other lower-limb angles were smaller, a consistent pattern was still present: the 10° FPA condition amplified the changes observed for the 5° condition. These joints may not be intentionally adjusted, but are instead influenced passively by changes in proximal joints. The results support the hypothesis that consistent patterns of joint angle changes occur for different FPA alterations. These findings could help developing predictive models to estimate how the mTFCF will change for different FPAs, based only on natural gait.

Pelvic rotation displayed condition-specific behavior: toe-in gait involved a contralateral transverse pelvic rotation, whereas toe-out gait showed minimal change. This may suggest that achieving a toe-in FPA relies more heavily on pelvic adjustment, possibly because the range of motion for internal hip rotation is smaller than external rotation [45]. Additionally, the large SD observed for toe-in 10° may reflect that the amount of transverse pelvic rotation required varies with individual hip flexibility. An increase in hip abduction was also observed during toe-out gait, which may reflect an adaptation to a wider step width. Research has shown that a wider step reduces peak KAM [46]. It should therefore be noted that results for toe-out gait might not only be influenced by the FPA, but also by associated gait modifications such as step width.

4.2. Changes in Medial Compartment Loading

The hypothesis that individuals would respond better to either a toe-in or toe-out FPA alteration for each peak metric was not supported. No clear pattern was observed where participants who reduced the mTFCF with toe-in showed increases with toe-out, or vice versa, across outcome measures. Instead, responses were heterogeneous. These findings indicate that the effect of different FPAs on the change in mTFCF is highly individual. A possible explanation is the difference in muscle activation patterns, which have a strong influence on internal knee loading [47]. Different FPAs may require different muscle forces, and for larger alterations, participants may adopt compensatory strategies to achieve the target FPA, potentially resulting in unnatural muscle activations. This study is, to the best of current knowledge, the first to investigate individual responses to multiple FPA alterations, therefore direct comparison with previous work is not possible. However, existing research already emphasizes the importance of a personalized approach when selecting a target FPA, and the results of this study further support the need for personalized targets based on each participant's response [17].

Based on reductions in first peak mTFCF, seven participants (70% of the responsive group) were assigned a target FPA of toe-in 10°. This is similar to previous KAM studies, where the majority of responsive participants were also prescribed a toe-in 10° target [48, 6]. When adopting the target FPA, the first peak and impulse mTFCF showed significant reductions, consistent with the set hypothesis. Interestingly, the second peak also decreased. First peak and impulse %tTFCF were also significantly reduced, whereas the second peak showed a slight increase, indicating that the medial compartment carried a greater proportion of the total load at that point in stance. Although the effects of a personalized target FPA on mTFCF have not been previously investigated, previous studies have examined the effects on the KAM. One study set the target FPA based on each participant's largest KAM peak, which was the first peak for most participants. They reported significant reductions in the first peak and impulse, but not in the second peak, indicating that second peak reductions were generally not expected [33]. Overall, these findings suggest that a personalized target FPA reduces medial compartment loading across the stance phase. The reduction in impulse is also encouraging, since both the first and second peak influence this parameter. As patients with knee OA have a higher loading in the first half of the stance phase compared to healthy individuals, but also walk with a greater percentage of total load on the medial compartment, it may be clinically useful to aim to reduce both the mTFCF and %tTFCF [49, 50].

4.3. Relationship Between ΔKAM and ΔmTFCF

The relationship between ΔKAM and ΔmTFCF was positive across all FPA conditions. The presence of a positive, but not perfect association is consistent with previous findings showing that KAM reflects only part of the mechanisms contributing to medial compartment loading [26]. The strength of this relationship differed between gait phases within each FPA condition, meaning that the associations observed for the first and second peak were not uniform. These phase-specific differences are also consistent with literature, further suggesting that the relation between KAM and medial compartment loading varies across the stance phase [51].

The strength of the association however also varied between FPA conditions, and the hypothesis that first peak changes in mTFCF would show stronger associations than for second peak across all FPA conditions was therefore not supported. For the first peak, changes in mTFCF were more strongly associated with changes in KAM for toe-out gait than for toe-in gait. This suggests that although additional factors other than the KAM influence mTFCF for both gait types, their contribution is smaller for toe-out gait during early stance. It also indicates that a given change in KAM corresponds to a larger change in mTFCF for toe-out compared with toe-in gait. In contrast, for the second peak, the pattern was reversed, with toe-in gait showing a stronger association than toe-out gait. These results suggest that the association depends on both gait phase and the direction of the FPA alteration.

Richards [28] reported a stronger relationship between changes in KAM and changes in mTFCF during toe-in gait for the first peak. Therefore, the opposite pattern observed in the present study was unexpected. Although this discrepancy cannot be attributed directly to one specific factor, several differences between the studies may help explain it. First, [28] studied patients with knee OA, whereas the present study examined healthy adults. Muscle weakness, particularly of the quadriceps and hamstrings, is commonly associated with knee OA [52]. Patients with knee OA also walk with increased muscle co-contraction compared to healthy individuals, to compensate for this weakness and joint instability [53, 54, 55]. Because muscle forces strongly influence knee joint loading, these differences in participant populations may lead to different loading responses to FPA alterations. Second, mTFCF estimation methods differed: [28] used a different musculoskeletal model and approach to determine how the tTFCF was distributed between the medial and lateral compartment. Finally, the toe-in FPA alteration in [28] was participant-specific, aiming for a 10% reduction in first peak KAM, while the present study applied fixed FPA alterations. As a result, KAM reductions were relatively uniform in [28] but varied across participants in the present study, affecting the observed associations.

For the impulse, no consistent differences in the relationship between ΔKAM and ΔmTFCF were observed between toe-in and toe-out conditions. This indicates that for the change in mTFCF over the whole stance phase, the relationship is less sensitive to the direction of FPA alteration. Additionally, R^2 values were generally higher for 10° conditions than for 5° conditions, suggesting that the association between ΔKAM and ΔmTFCF may be more consistent with larger FPA alterations. However, this pattern could also reflect other factors, such as the influence of measurement noise. Smaller changes are more susceptible to noise, which could weaken the observed relationship.

4.4. Limitations

The musculoskeletal modeling methods used in this study have several limitations. The knee was represented as a single degree-of-freedom joint, capturing only flexion-extension [39]. The knee abduction-adduction and internal-external rotation could therefore not be captured. These joint angles are most likely also affected by changes in FPA, since modifying the FPA is caused by adjustments at the hip, which in turn alter the kinematic chain at the knee. Research has also shown that the knee adduction angle influences the medial compartment load and medial to lateral force ratio [56]. Therefore, changes in these kinematics remain uncharacterized, and any effects of knee adduction on the mTFCF were not captured. However, this musculoskeletal model was chosen because it is commonly used in research for estimating knee joint loading in the medial and lateral compartment, and the model has been validated for other gait modifications (medial knee thrust and lateral trunk lean), showing that it is able to capture differences caused by gait patterns [27]. Future research could apply more advanced frameworks, such as OpenSim Joint Articular Mechanics (JAM) [57], which allows a six degree-of-freedom representation of the knee, and representations of articular contact and passive ligament structures.

The musculoskeletal model assumes that the tibiofemoral contact force occurs at a single point within each compartment [29]. Subject-specific parameters, such as frontal-plane alignment (i.e. knee varus-valgus) and medial-lateral contact locations, therefore have a strong influence on the estimated contact forces. Lerner [29] reported that small errors in both alignment and contact locations can lead to large errors in predicted peak loads. Because radiographic images where not available for the present study, these parameters could not be personalized. Therefore, the predicted mTFCF may not fully reflect individual knee mechanics.

Muscle forces were determined using a cost function that minimizes muscle activation, and EMG data were not collected. As a result, it is unclear how accurately the estimated muscle forces reflect each participant's neuromuscular strategy. This approach may even be less accurate for patients with knee OA, who often exhibit greater co-contraction and altered activation patterns. Previous work has also shown that muscle force adaptations to gait retraining vary substantially across knee OA patients [58]. Because muscle forces strongly influence estimated mTFCF [47], this method may limit the accuracy of predicted changes in knee loading in response to FPA alterations.

Participant characteristics limit the generalizability of this study, as only healthy participants were included. Aside from the differences in neuromuscular strategies, the BMI was also lower than typically observed in knee OA patients. A higher BMI is associated with greater peak compressive knee forces [59]. Therefore, although the age was similar to the target OA population [60], the effects of FPA alterations on medial compartment loading observed here may not generalize to the patient population.

For three FPA conditions (second peak toe-in 10° and impulse toe-in and toe-out 5°), one assumption of linear regression, either normality or homoscedasticity of residuals, was violated. Consequently, the strength and slope of the reported associations for these conditions should be interpreted with caution, as they may not fully represent the relationship between ΔKAM and ΔmTFCF.

Lastly, because this study used a cross-sectional design, it only captures the immediate effects of a target FPA. Previous research has shown that reductions in KAM following a personalized target FPA were only partially retained after one year [6]. It is therefore possible that a similar pattern could occur for mTFCF, meaning that the reductions observed in the current study might change over time when continuing to walk with an altered FPA.

5 Conclusion

This study demonstrated that lower-limb joint angles were consistently altered for different FPAs. Toe-in and toe-out conditions produced changes in opposite directions, and the magnitude of these changes increased with larger FPA alterations. The largest differences were observed for hip rotation and subtalar inversion/eversion. The change in mTFCF for toe-in and toe-out FPA alterations was highly individual, with no consistent directional pattern across participants for each peak metric, emphasizing the need for personalized targets. When using a personalized target FPA based on the first peak mTFCF, significant reductions were observed in the first peak, second peak and impulse mTFCF, reducing medial compartment loading across the stance phase. The relationship between changes in KAM and mTFCF was positive for all conditions, but the strength depended on both stance phase and the direction of the FPA alteration.

Future studies should focus on patients with knee OA to investigate how FPA alterations influence medial compartment loading in the target population. Additionally, future research should incorporate individually adapted musculoskeletal models that account for subject-specific lower-limb alignment and neuromuscular control, to improve the accuracy of medial compartment loading estimates.

A
WalkWell Gait reTrainer

This study is part of the project *WalkWell Gait reTrainer: Remote Patient Monitoring and Gait Retraining for Knee Osteoarthritis* [34]. This project is a collaboration between the Erasmus MC, TU Delft and Moveshelf.

In most studies investigating the foot progression angle (FPA), participants are required to visit a motion capture (MoCap) laboratory on a weekly basis to practice walking with their target FPA, while being instructed to practice at home without feedback. Because MoCap systems are costly and not widely available in clinical settings, and given the large population of knee osteoarthritis patients, this approach is not feasible. Consequently, there is growing interest in simpler systems that are able to measure the FPA at home, such as the WalkWell device.

The WalkWell device consists of an inertial measurement unit sensor (IMU), placed on the foot of the user. The IMU measures orientation, change in orientation and velocity during gait. The IMU is connected via Bluetooth to a mobile phone where the app Moveshelf Mobile is installed. Within the app, the raw IMU data is fed into an algorithm, and the FPA per step is calculated in real-time [36]. Based on a personal target FPA, the user is given visual and audio feedback while walking.

For the parent study, healthy participants attend a MoCap laboratory once where baseline measurements are conducted, as described in the present study. After receiving a target FPA based on the knee adduction moment, participants are asked to practice walking with the target FPA while using the device, at least three times during one week.

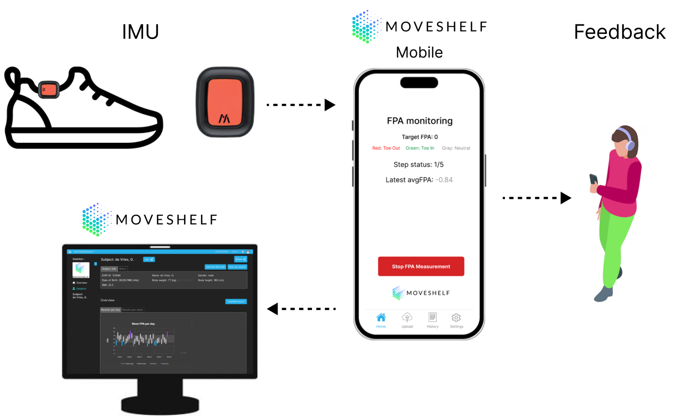
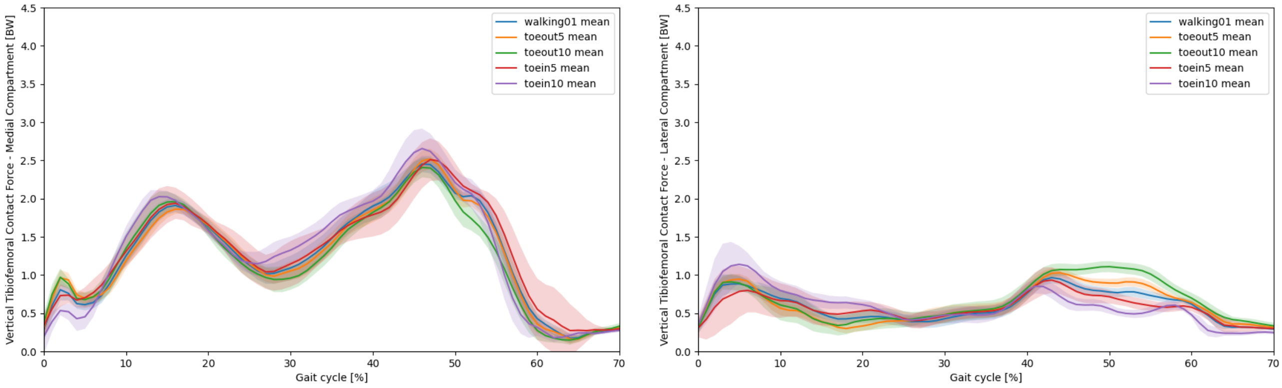


Figure A.1. Overview of the WalkWell Gait reTrainer. The inertial measurement unit (IMU) is placed on the shoe. Measurements from IMU are sent to the Moveshelf Mobile app, where the foot progression angle (FPA) is calculated. The user receives visual and audio feedback on their FPA. The therapist can access training data through the Moveshelf platform.

B

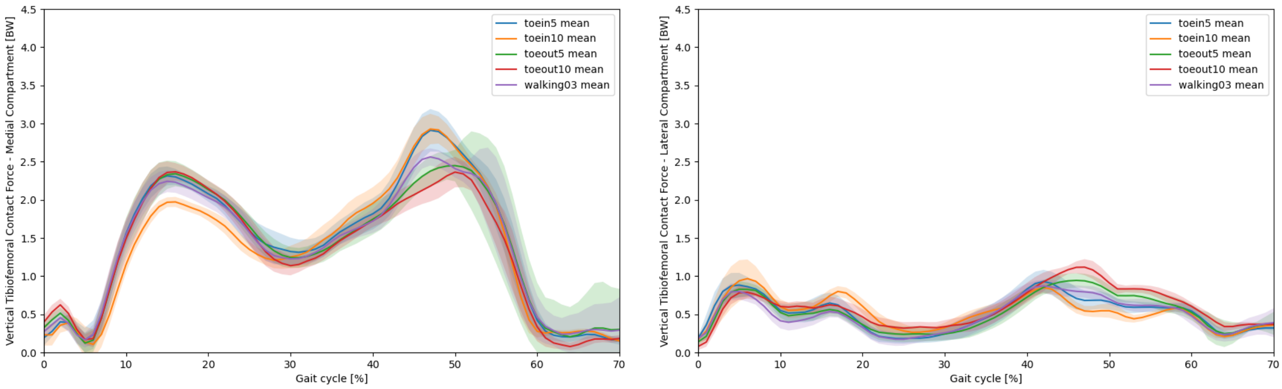
Tibiofemoral Contact Forces

Estimated tibiofemoral contact forces, for each participant for each condition. Only the stance phase of the gait cycle is plotted. The shaded band around the mean represents the mean ± standard deviation.



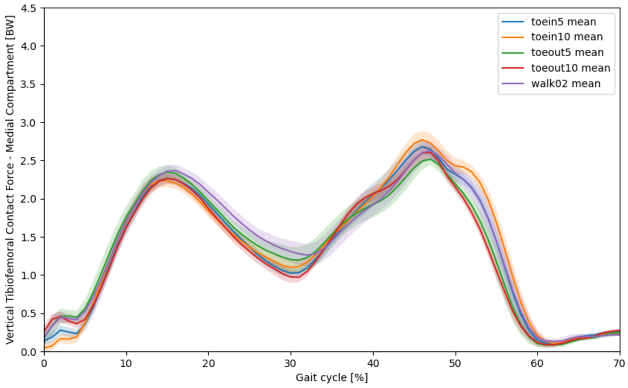
(a) Medial TFCF **(b)** Lateral TFCF

Figure B.1. Estimated tibiofemoral contact forces for each condition (subject 1).

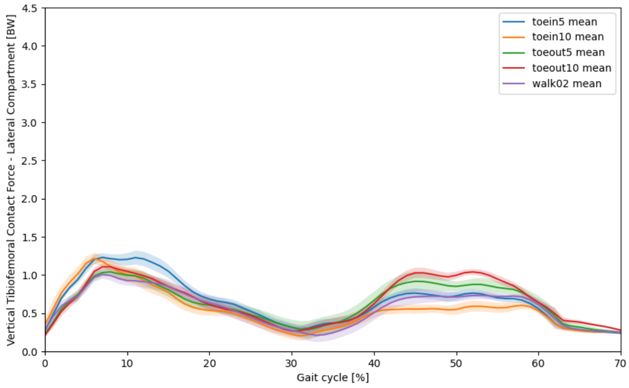


(a) Medial TFCF **(b)** Lateral TFCF

Figure B.2. Estimated tibiofemoral contact forces for each condition (subject 2).

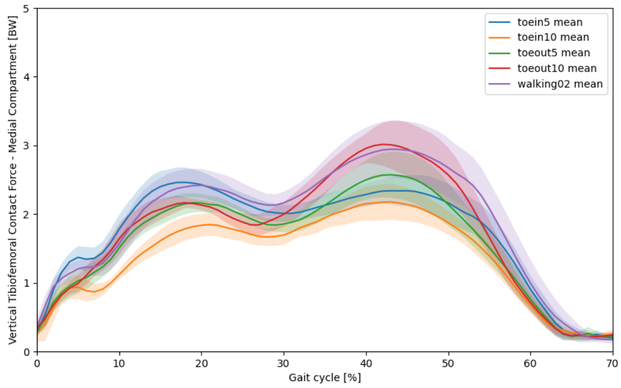


(a) Medial TFCF

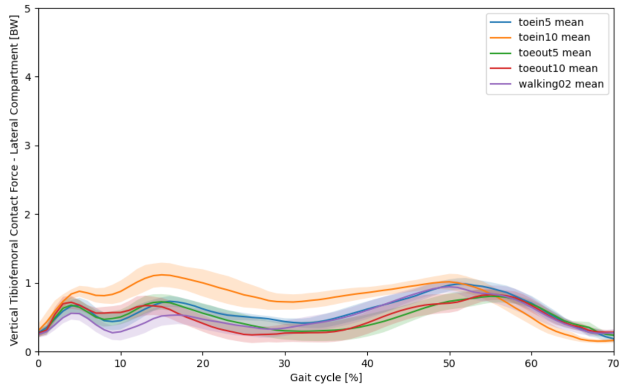


(b) Lateral TFCF

Figure B.3. Estimated tibiofemoral contact forces for each condition (subject 3).

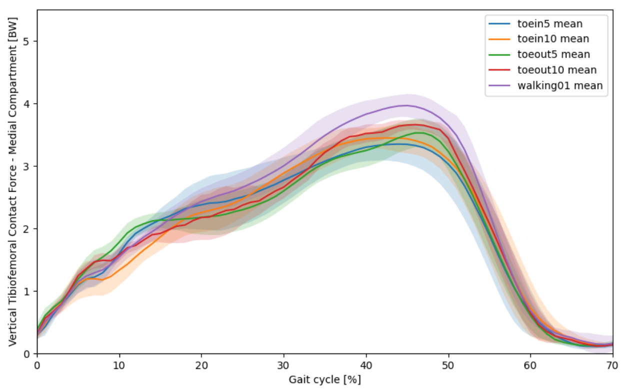


(a) Medial TFCF

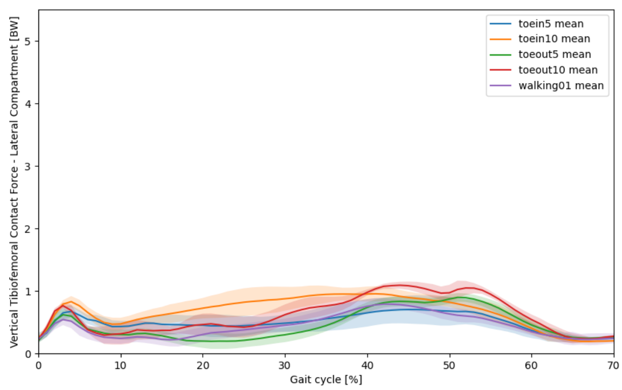


(b) Lateral TFCF

Figure B.4. Estimated tibiofemoral contact forces for each condition (subject 4).

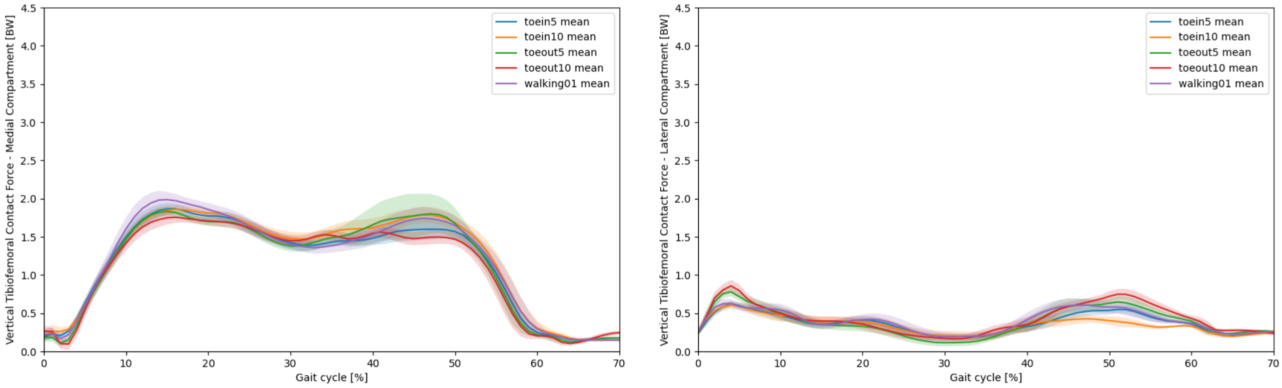


(a) Medial TFCF



(b) Lateral TFCF

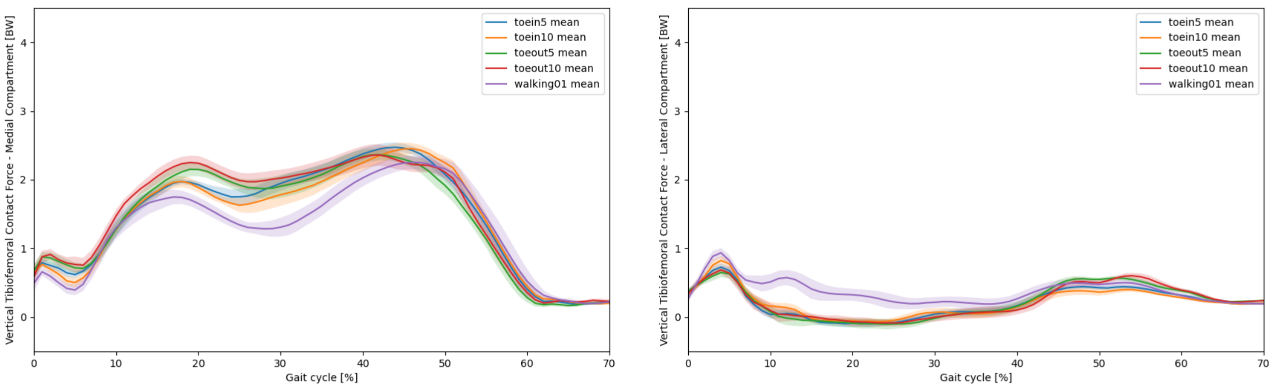
Figure B.5. Estimated tibiofemoral contact forces for each condition (subject 5).



(a) Medial TFCF

(b) Lateral TFCF

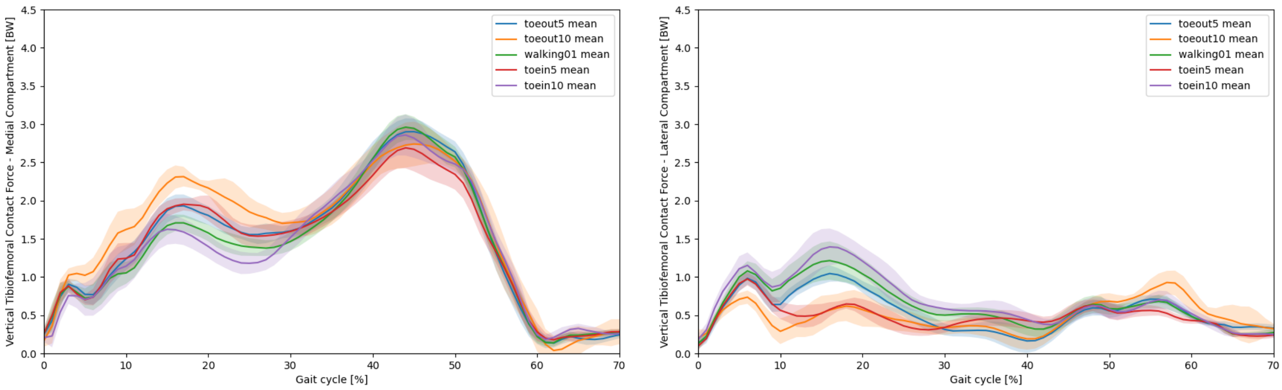
Figure B.6. Estimated tibiofemoral contact forces for each condition (subject 6).



(a) Medial TFCF

(b) Lateral TFCF

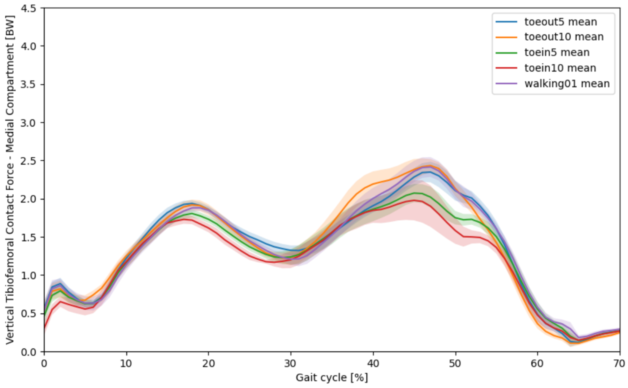
Figure B.7. Estimated tibiofemoral contact forces for each condition (subject 7).



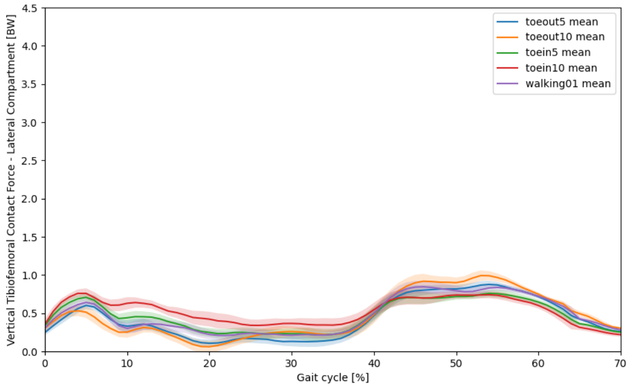
(a) Medial TFCF

(b) Lateral TFCF

Figure B.8. Estimated tibiofemoral contact forces for each condition (subject 8).

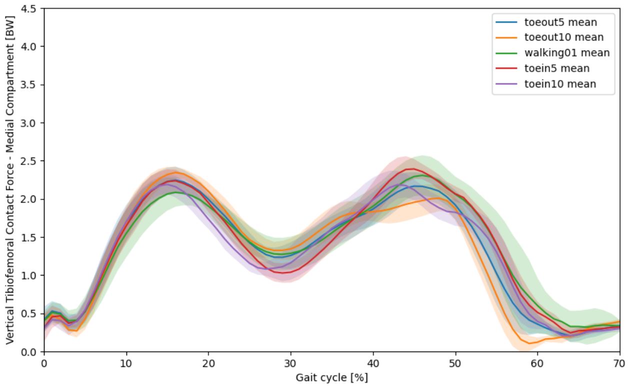


(a) Medial TFCF

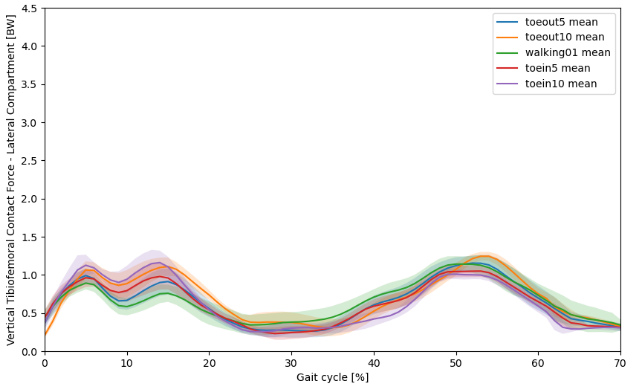


(b) Lateral TFCF

Figure B.9. Estimated tibiofemoral contact forces for each condition (subject 9).

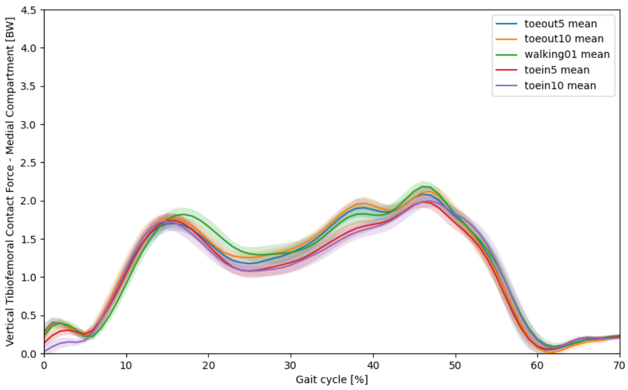


(a) Medial TFCF

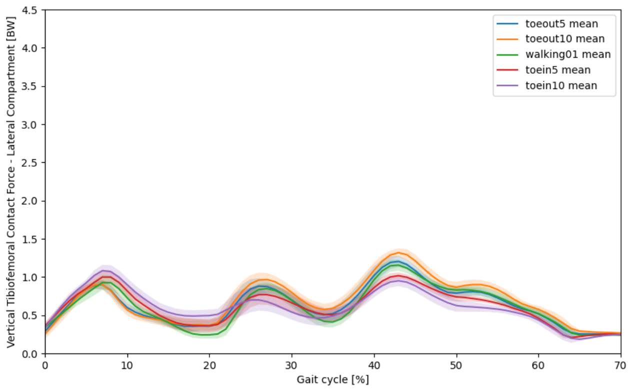


(b) Lateral TFCF

Figure B.10. Estimated tibiofemoral contact forces for each condition (subject 10).

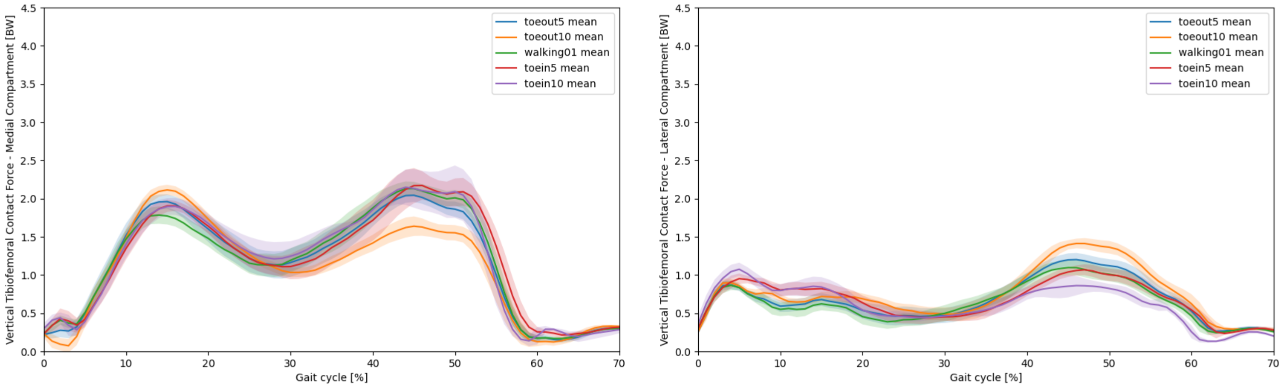


(a) Medial TFCF



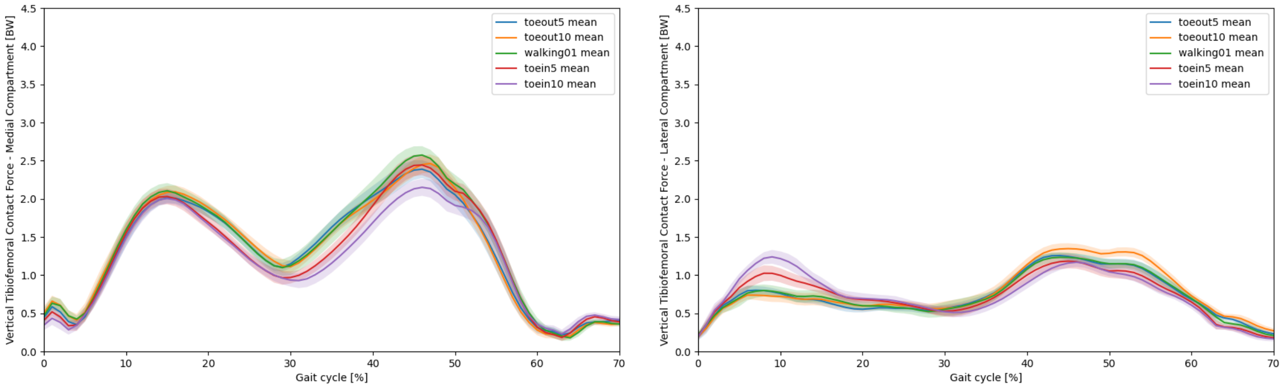
(b) Lateral TFCF

Figure B.11. Estimated tibiofemoral contact forces for each condition (subject 11).



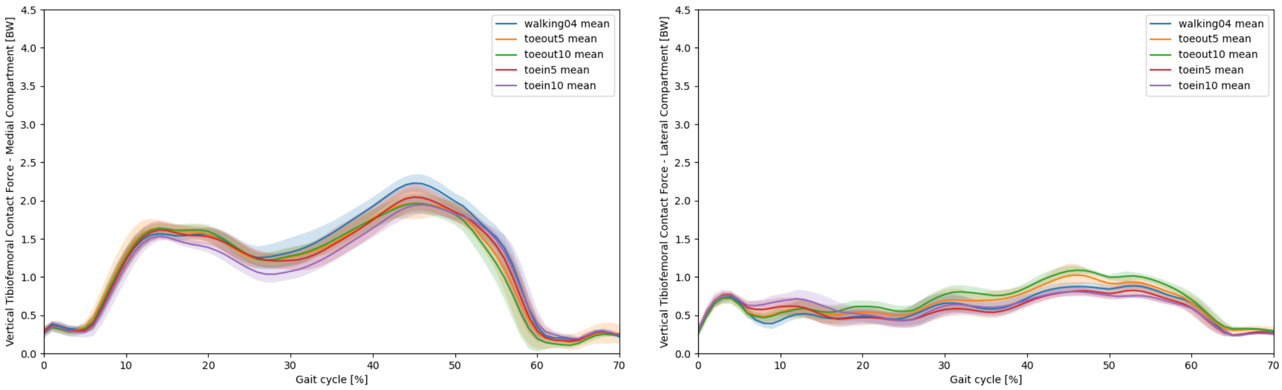
(a) Medial TFCF **(b)** Lateral TFCF

Figure B.12. Estimated tibiofemoral contact forces for each condition (subject 12).



(a) Medial TFCF **(b)** Lateral TFCF

Figure B.13. Estimated tibiofemoral contact forces for each condition (subject 13).



(a) Medial TFCF **(b)** Lateral TFCF

Figure B.14. Estimated tibiofemoral contact forces for each condition (subject 14).

C

Knee Adduction Moments

Knee adduction moment obtained from ground reaction forces and marker data, for each participant for each condition. Only the stance phase of the gait cycle is plotted. The shaded band around the mean represents the mean ± standard deviation.

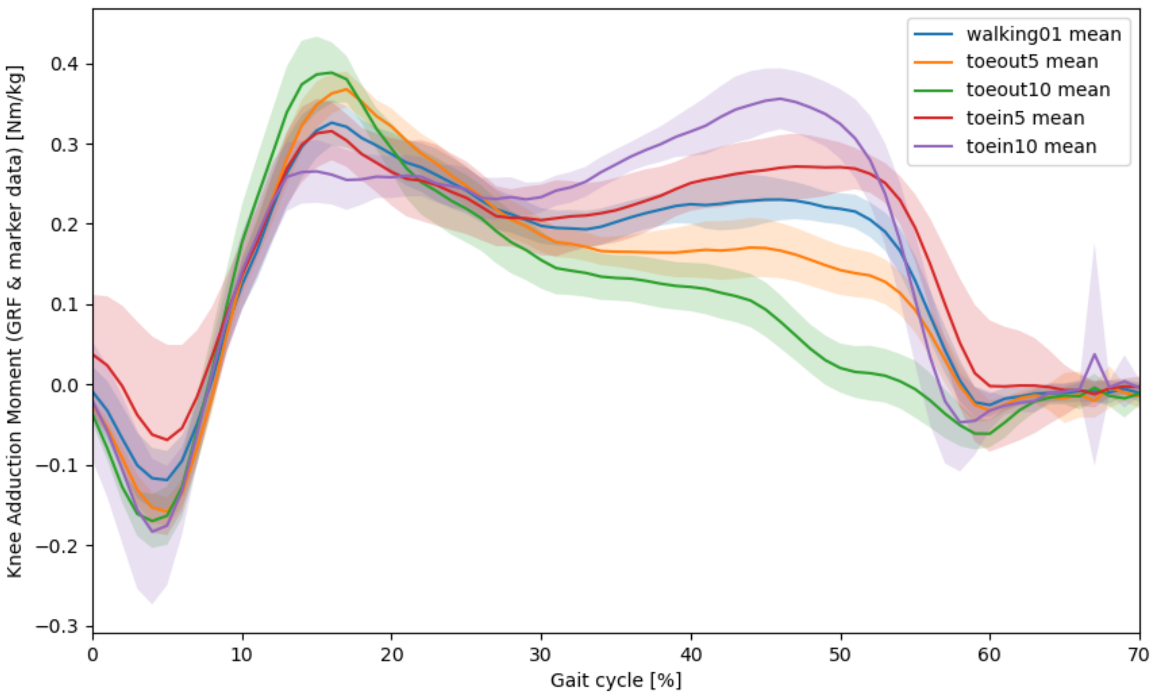


Figure C.1. Knee Adduction Moment for each condition (subject 1).

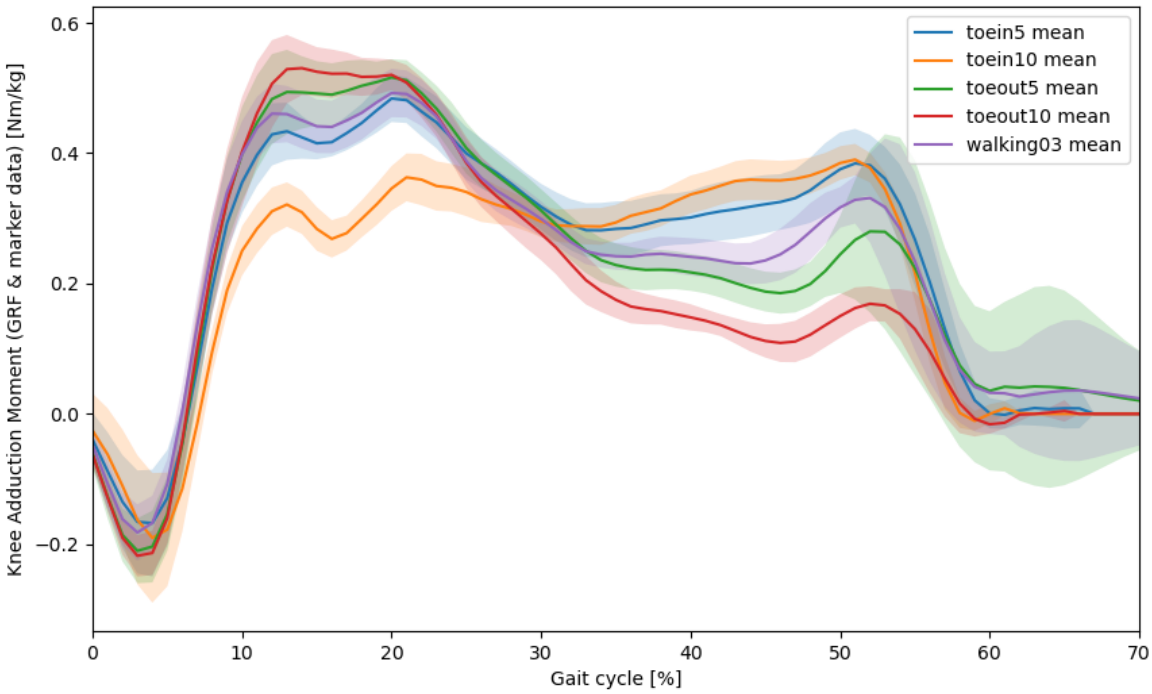


Figure C.2. Knee adduction moment for each condition (subject 2).

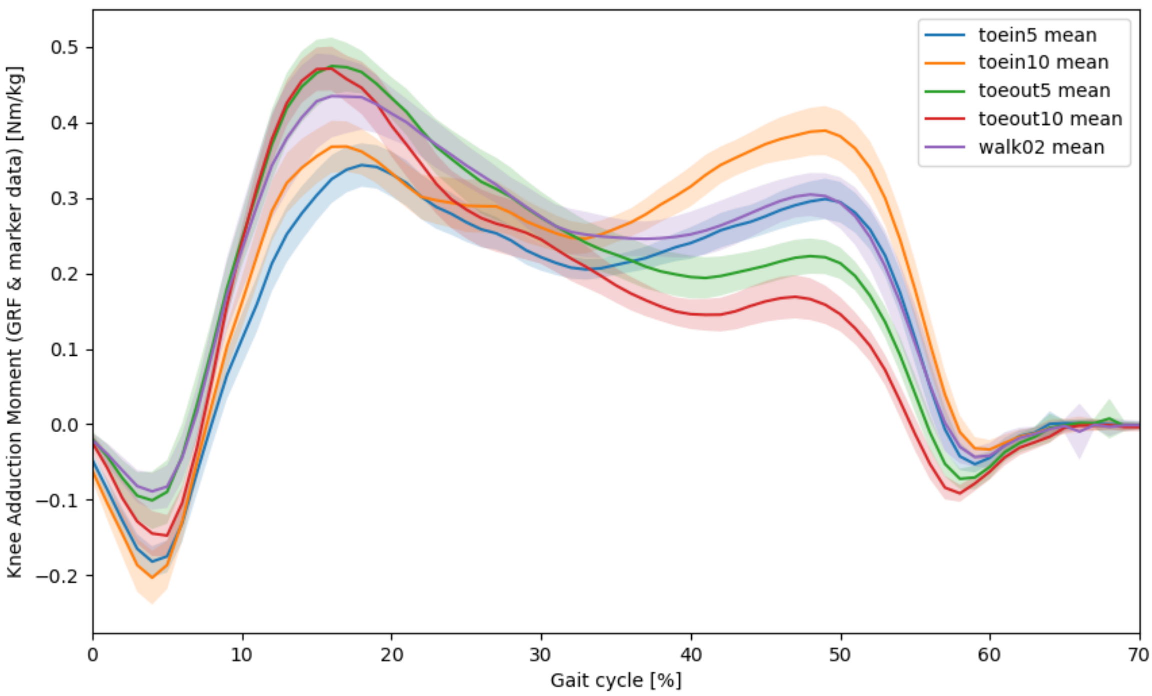


Figure C.3. Knee adduction moment for each condition (subject 3).

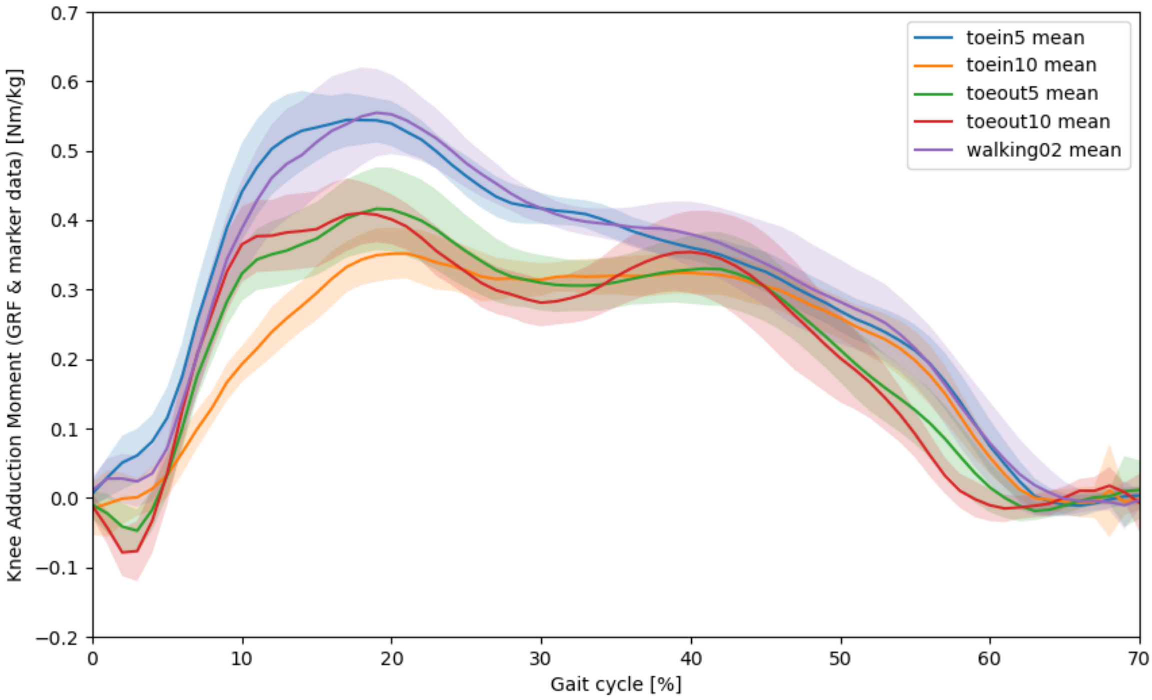


Figure C.4. Knee adduction moment for each condition (subject 4).

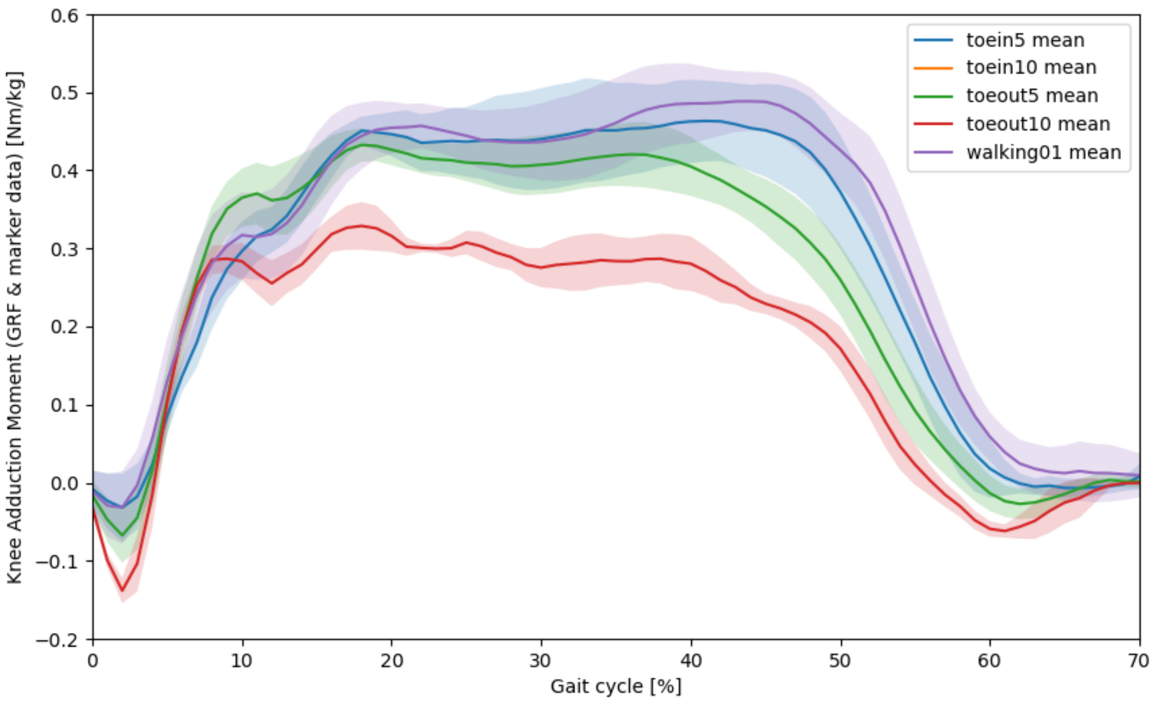


Figure C.5. Knee adduction moment for each condition (subject 5).

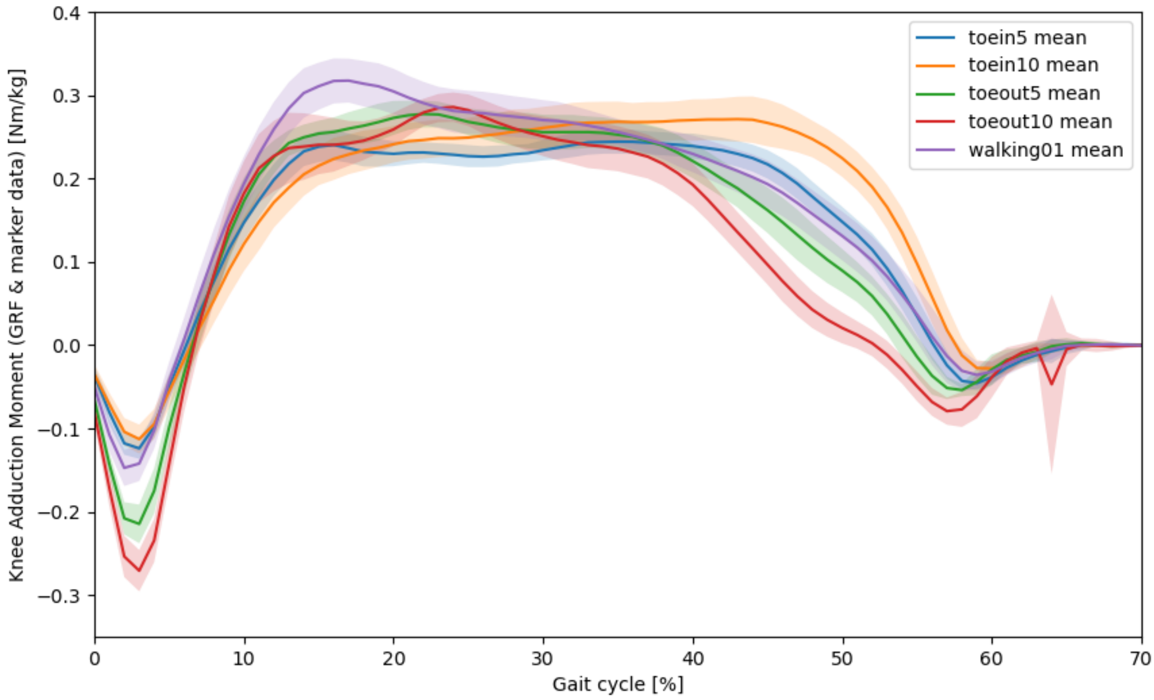


Figure C.6. Knee adduction moment for each condition (subject 6).

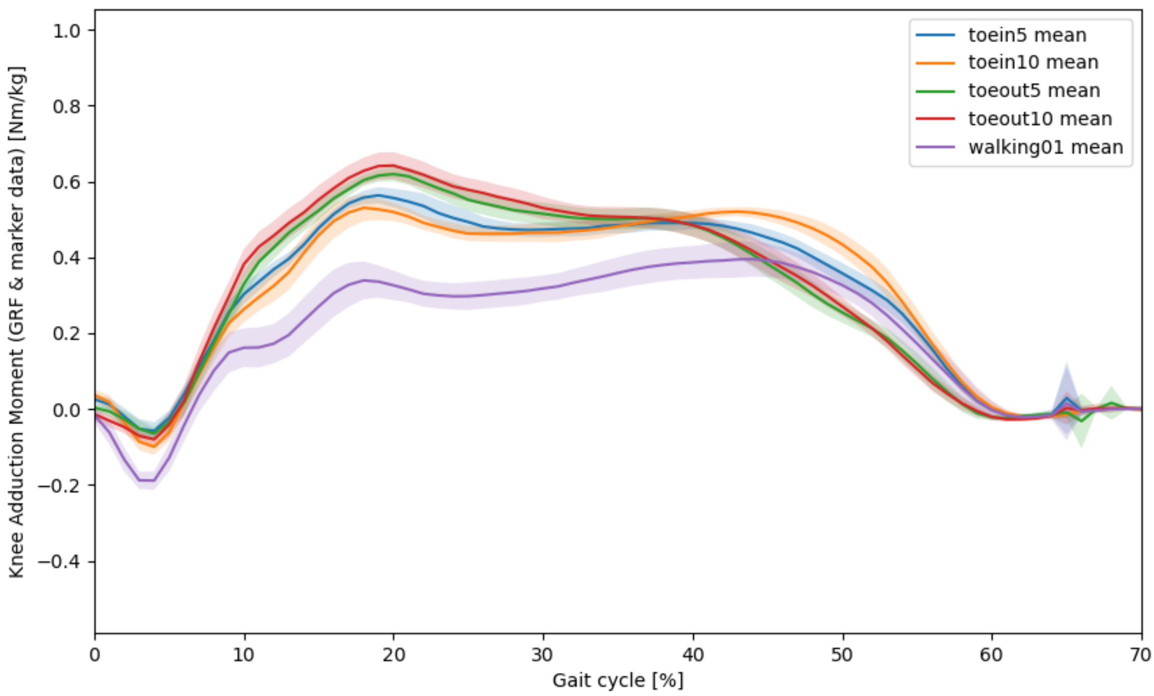


Figure C.7. Knee adduction moment for each condition (subject 7).

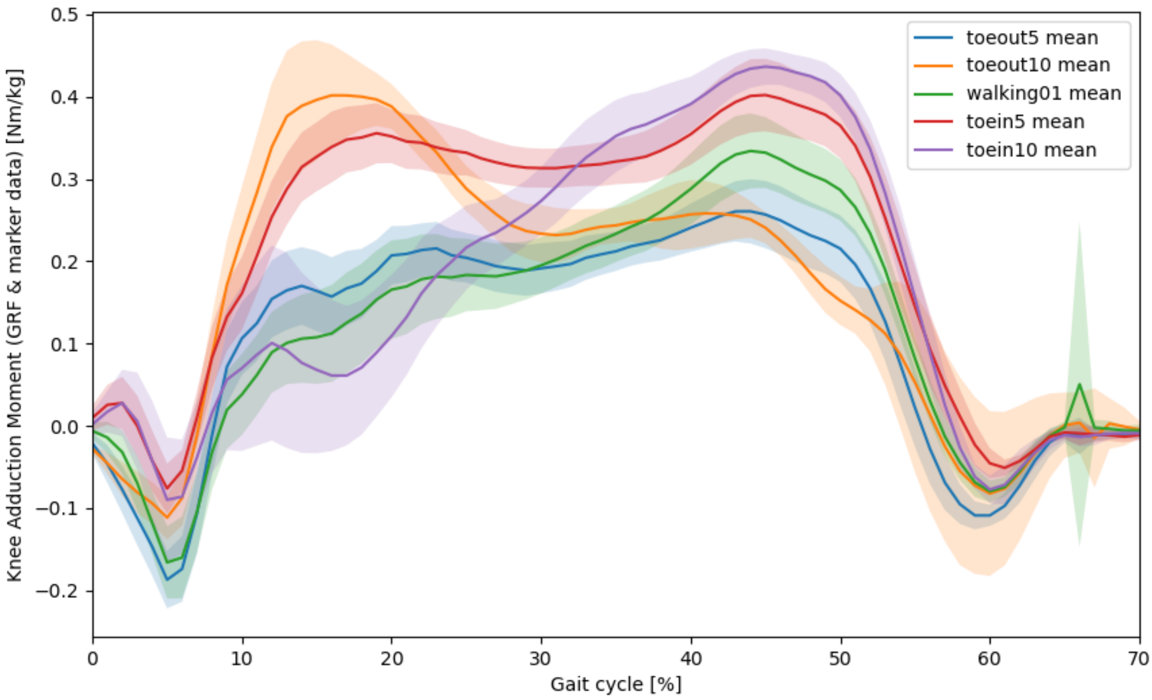


Figure C.8. Knee adduction moment for each condition (subject 8).

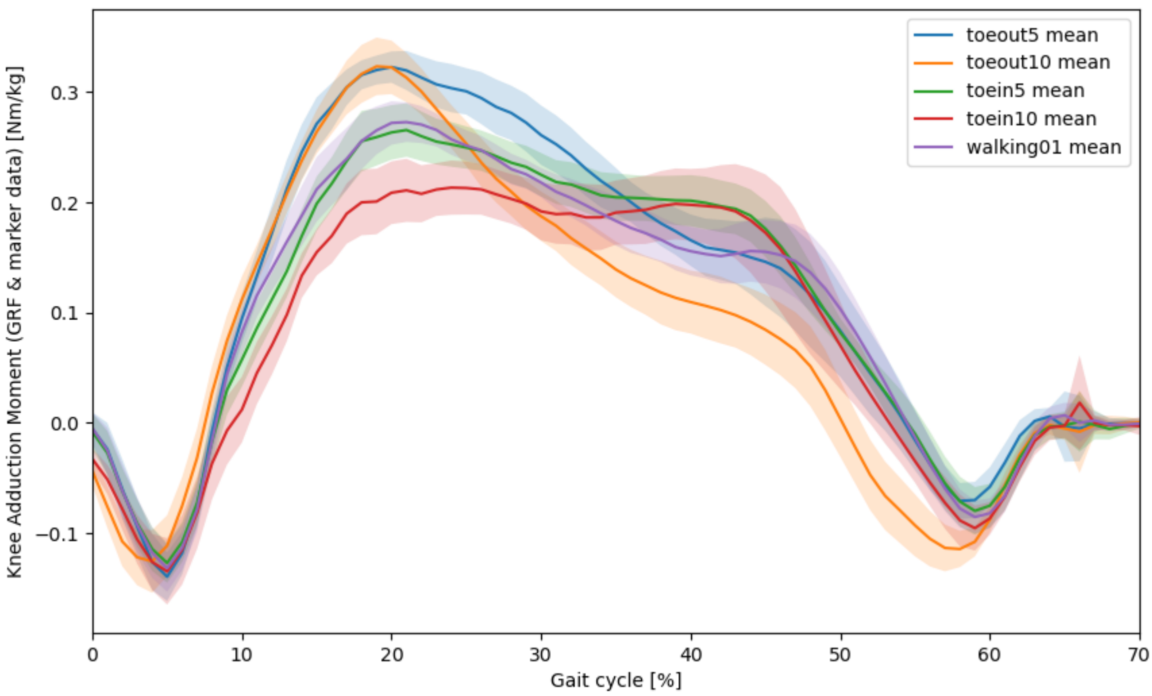


Figure C.9. Knee adduction moment for each condition (subject 9).

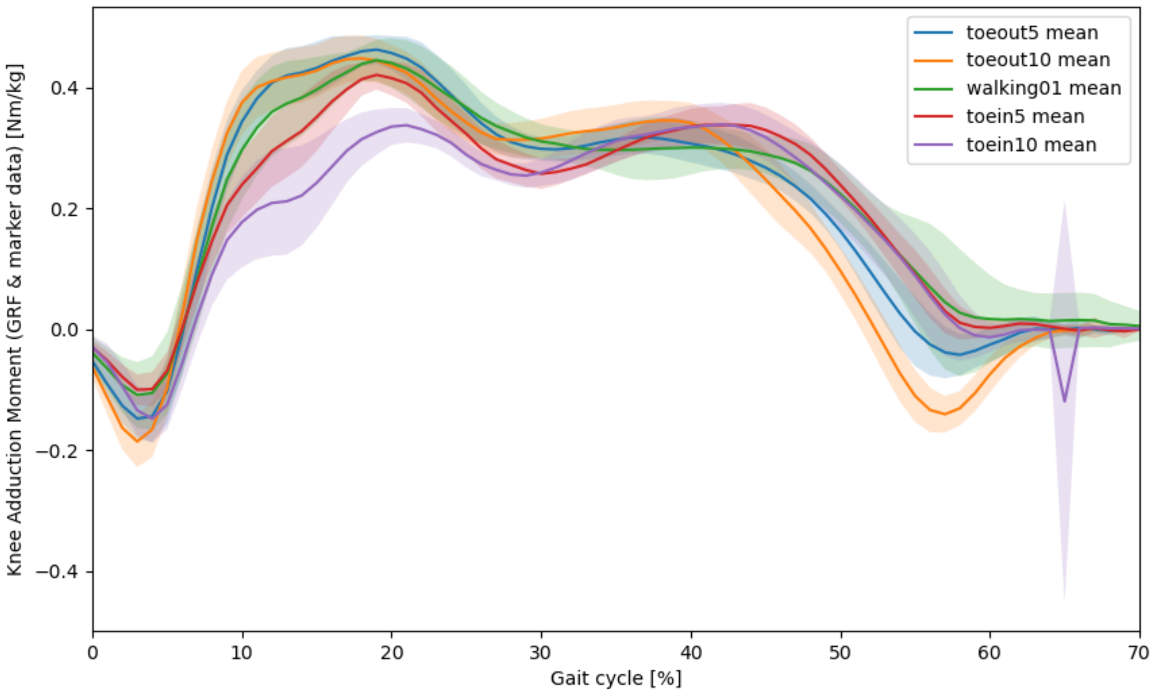


Figure C.10. Knee adduction moment for each condition (subject 10).

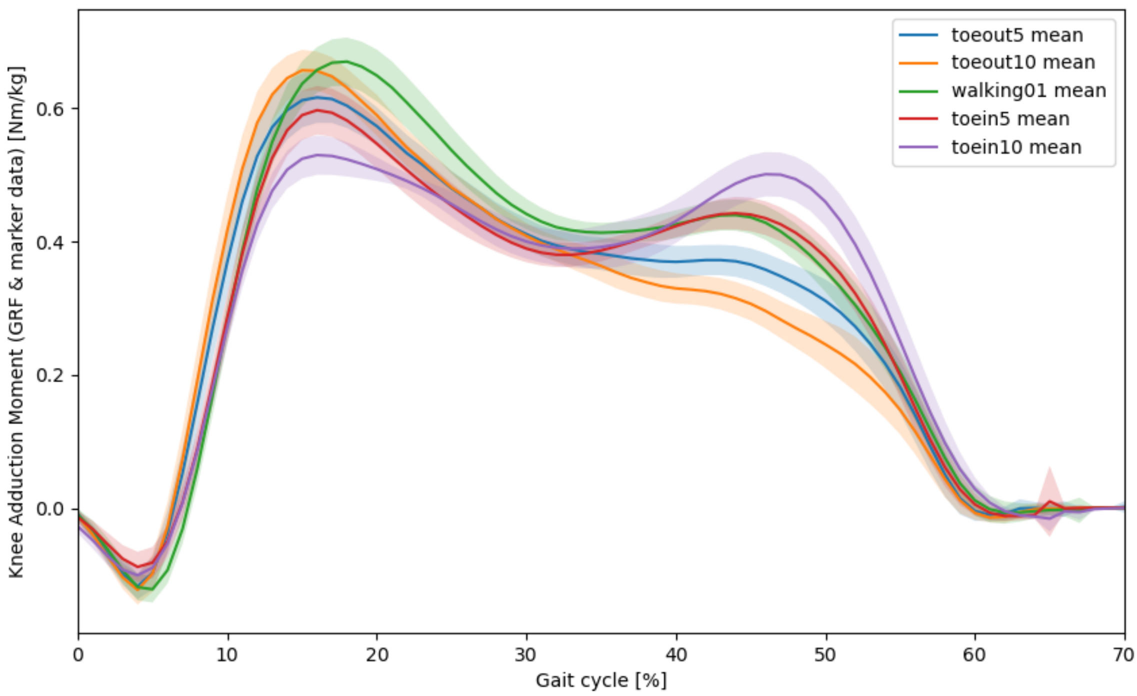


Figure C.11. Knee adduction moment for each condition (subject 11).

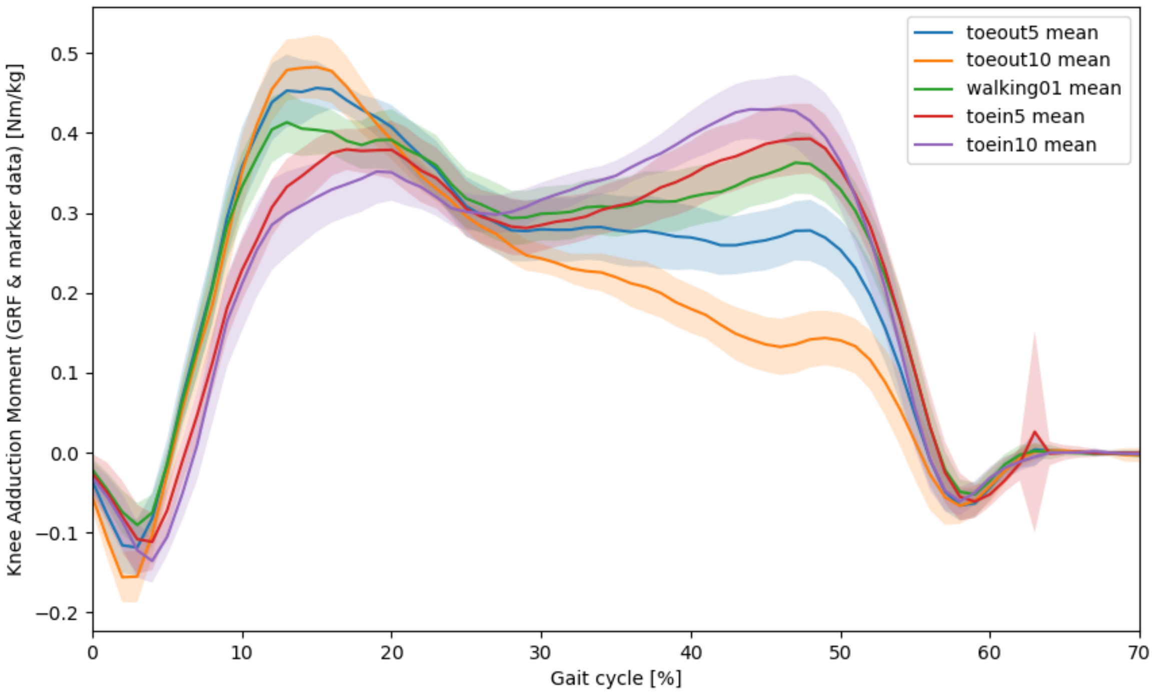


Figure C.12. Knee adduction moment for each condition (subject 12).

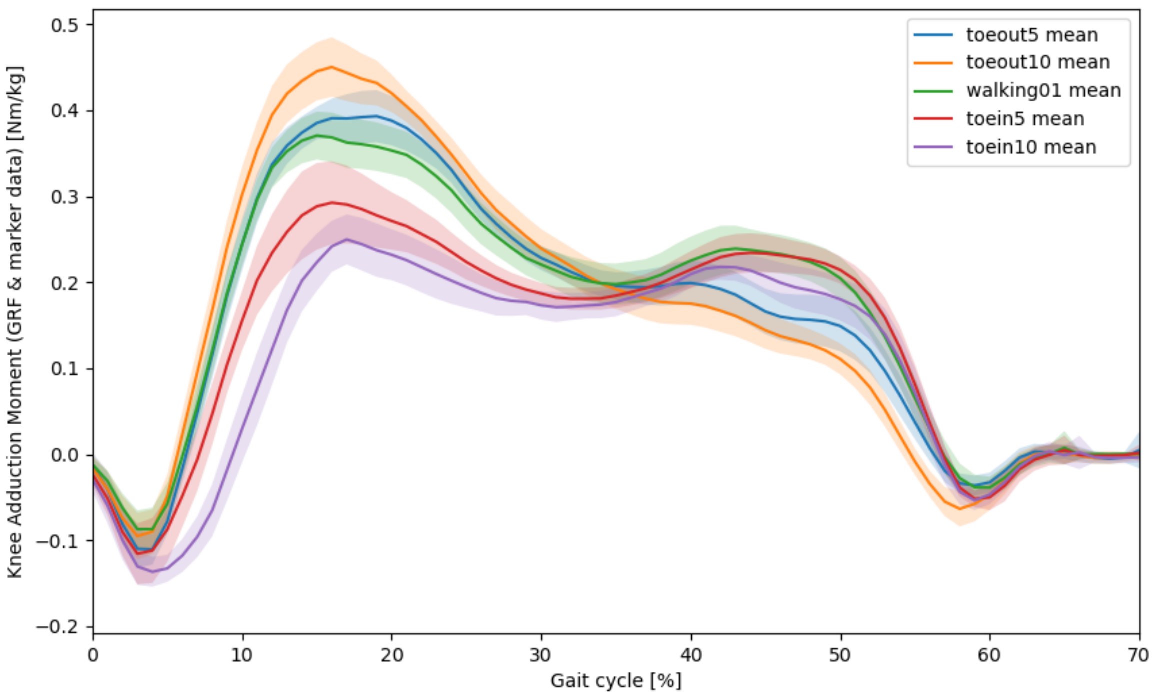


Figure C.13. Knee adduction moment for each condition (subject 13).

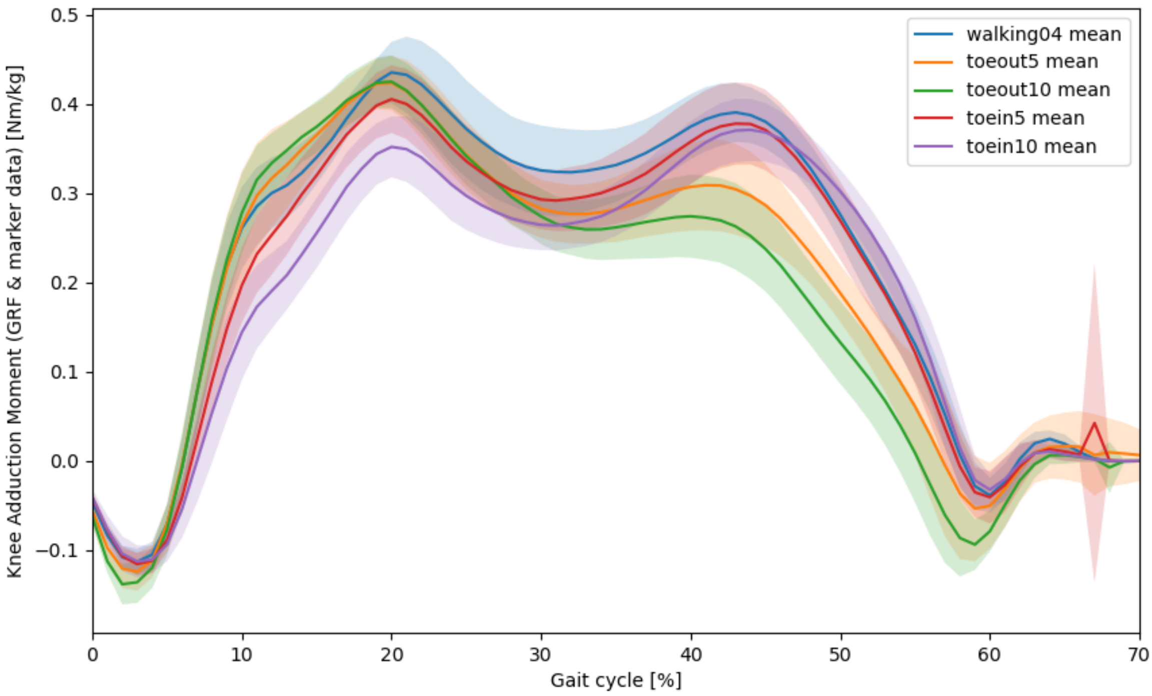


Figure C.14. Knee adduction moment for each condition (subject 14).

D

Results Target FPA

Table D.1. Results for the change in mTFCF (ΔmTFCF) and change in %tTFCF (Δ%tTFCF) parameters for the defined target FPA.

Target FPA	Participant Number	ΔmTFCF [BW]			Δ%tTFCF [%]		
		First Peak	Second Peak	Impulse	First Peak	Second Peak	Impulse
Toe-out 5°	1	-0.044	0.048	-0.035	4.120	-1.927	-0.759
	5	-0.454	-0.444	-0.329	2.396	-3.008	-1.990
Toe-out 10°	6	-0.228	-0.196	-0.155	-1.866	5.324	-2.003
Toe-in 10°	2	-0.275	0.248	-0.084	-8.005	6.810	-2.314
	3	-0.137	0.116	-0.023	-0.649	4.879	0.558
	4	-0.625	-0.817	-0.432	-14.91	-9.376	-13.77
	9	-0.153	-0.420	-0.114	-9.008	1.034	-4.549
	11	-0.141	-0.161	-0.059	-9.856	3.055	-1.325
	13	-0.099	-0.426	-0.144	-5.188	-2.935	-3.652
	14	-0.067	-0.253	-0.126	-6.714	-0.461	-2.323

Abbreviations. FPA = foot progression angle, ΔmTFCF = difference in medial tibiofemoral contact forces between the target FPA condition and natural walking, Δ%tTFCF = difference in medial tibiofemoral contact forces expressed as percentage of the total tibiofemoral contact force between the target FPA condition and natural walking.
The target FPA was defined as the condition that lowered the first peak mTFCF the most. Participants 7, 8, 10 and 12 did not improve their first peak mTFCF with any of the FPA conditions.

Bibliography

[1] Huibin Long et al. “Prevalence trends of Site-Specific osteoarthritis from 1990 to 2019: Findings from the Global Burden of Disease Study 2019”. In: *Arthritis & Rheumatology* 74.7 (Mar. 2022), pp. 1172–1183. DOI: 10.1002/art.42089. URL: https://doi.org/10.1002/art.42089.

[2] H. Hsu and R. M. Siwiec. “Knee Osteoarthritis”. In: *StatPearls [Internet]*. Updated 2025 Jan–. PMID: 29939661. Treasure Island (FL): StatPearls Publishing, June 2023.

[3] David J. Hunter and Sita Bierma-Zeinstra. “Osteoarthritis”. In: *The Lancet* 393.10182 (Apr. 2019), pp. 1745–1759. DOI: 10.1016/s0140-6736(19)30417-9. URL: https://doi.org/10.1016/s0140-6736(19)30417-9.

[4] Paul Dieppe and Kenneth D. Brandt. “What is important in treating osteoarthritis? Whom should we treat and how should we treat them?” In: *Rheumatic Disease Clinics of North America* 29.4 (Nov. 2003), pp. 687–716. DOI: 10.1016/s0889-857x(03)00054-1. URL: https://doi.org/10.1016/s0889-857x(03)00054-1.

[5] Barton L. Wise et al. “Patterns of compartment involvement in tibiofemoral osteoarthritis in men and women and in whites and African Americans”. In: *Arthritis Care & Research* 64.6 (Jan. 2012), pp. 847–852. DOI: 10.1002/acr.21606. URL: https://doi.org/10.1002/acr.21606.

[6] Scott D Uhlrich et al. “Personalised gait retraining for medial compartment knee osteoarthritis: a randomised controlled trial”. In: *The Lancet Rheumatology* 7.10 (Aug. 2025), e708–e718. DOI: 10.1016/s2665-9913(25)00151-1. URL: https://doi.org/10.1016/s2665-9913(25)00151-1.

[7] Rosie Richards et al. “Gait Retraining With Real-Time Biofeedback to Reduce Knee Adduction Moment: Systematic Review of Effects and Methods Used”. In: *Archives of Physical Medicine and Rehabilitation* 98.1 (2017), pp. 137–150. ISSN: 0003-9993. DOI: https://doi.org/10.1016/j.apmr.2016.07.006. URL: https://www.sciencedirect.com/science/article/pii/S0003999316303471.

[8] Rebecca Rynne et al. “Effectiveness of gait retraining interventions in individuals with hip or knee osteoarthritis: A systematic review and meta-analysis”. In: *Gait & Posture* 95 (2022), pp. 164–175. ISSN: 0966-6362. DOI: https://doi.org/10.1016/j.gaitpost.2022.04.013. URL: https://www.sciencedirect.com/science/article/pii/S0966636222001072.

[9] Yoann Dessery et al. “Comparison of three knee braces in the treatment of medial knee osteoarthritis”. In: *The Knee* 21.6 (July 2014), pp. 1107–1114. DOI: 10.1016/j.knee.2014.07.024. URL: https://doi.org/10.1016/j.knee.2014.07.024.

[10] Robert J. Butler et al. “The effect of a subject-specific amount of lateral wedge on knee mechanics in patients with medial knee osteoarthritis”. In: *Journal of Orthopaedic Research®* 25.9 (Apr. 2007), pp. 1121–1127. DOI: 10.1002/jor.20423. URL: https://doi.org/10.1002/jor.20423.

[11] Jennifer C. Erhart et al. “A variable-stiffness shoe lowers the knee adduction moment in subjects with symptoms of medial compartment knee osteoarthritis”. In: *Journal of Biomechanics* 41.12 (Aug. 2008), pp. 2720–2725. DOI: 10.1016/j.jbiomech.2008.06.016. URL: https://doi.org/10.1016/j.jbiomech.2008.06.016.

[12] Michael A. Hunt et al. “Feasibility of a gait retraining strategy for reducing knee joint loading: Increased trunk lean guided by real-time biofeedback”. In: *Journal of Biomechanics* 44.5 (Dec. 2010), pp. 943–947. DOI: 10.1016/j.jbiomech.2010.11.027. URL: https://doi.org/10.1016/j.jbiomech.2010.11.027.

[13] R.T.H. Cheung et al. “Immediate and short-term effects of gait retraining on the knee joint moments and symptoms in patients with early tibiofemoral joint osteoarthritis: a randomized controlled trial”. In: *Osteoarthritis and Cartilage* 26.11 (Aug. 2018), pp. 1479–1486. DOI: 10.1016/j.joca.2018.07.011. URL: https://doi.org/10.1016/j.joca.2018.07.011.

[14] R. Richards et al. “Gait retraining using real-time feedback in patients with medial knee osteoarthritis: Feasibility and effects of a six-week gait training program”. In: *The Knee* 25.5 (June 2018), pp. 814–824. DOI: 10.1016/j.knee.2018.05.014. URL: https://doi.org/10.1016/j.knee.2018.05.014.

[15] M. Simic et al. “Altering foot progression angle in people with medial knee osteoarthritis: the effects of varying toe-in and toe-out angles are mediated by pain and malalignment”. In: *Osteoarthritis and Cartilage* 21.9 (Aug. 2013), pp. 1272–1280. DOI: 10.1016/j.joca.2013.06.001. URL: https://doi.org/10.1016/j.joca.2013.06.001.

[16] Pete B. Shull et al. “Toe-in gait reduces the first peak knee adduction moment in patients with medial compartment knee osteoarthritis”. In: *Journal of Biomechanics* 46.1 (Nov. 2012), pp. 122–128. DOI: 10.1016/j.jbiomech.2012.10.019. URL: https://doi.org/10.1016/j.jbiomech.2012.10.019.

[17] Scott D. Uhlrich et al. “Subject-specific toe-in or toe-out gait modifications reduce the larger knee adduction moment peak more than a non-personalized approach”. In: *Journal of Biomechanics* 66 (Jan. 2018), pp. 103–110. DOI: 10.1016/j.jbiomech.2017.11.003. URL: https://doi.org/10.1016/j.jbiomech.2017.11.003.

[18] S. Wang et al. “Sensor-Based Gait Retraining Lowers Knee Adduction Moment and Improves Symptoms in Patients with Knee Osteoarthritis: A Randomized Controlled Trial”. In: *Sensors* 21.16 (Aug. 2021), p. 5596. DOI: 10.3390/s21165596. URL: https://doi.org/10.3390/s21165596.

[19] Pete B Shull et al. “Six-week gait retraining program reduces knee adduction moment, reduces pain, and improves function for individuals with medial compartment knee osteoarthritis”. In: *Journal of Orthopaedic Research* 31.7 (2013), pp. 1020–1025. URL: https://doi.org/10.1002/jor.22340.

[20] M.A. Hunt and J. Takacs. “Effects of a 10-week toe-out gait modification intervention in people with medial knee osteoarthritis: a pilot, feasibility study”. In: *Osteoarthritis and Cartilage* 22.7 (May 2014), pp. 904–911. DOI: 10.1016/j.joca.2014.04.007. URL: https://doi.org/10.1016/j.joca.2014.04.007.

[21] Benjamin J. Fregly, Jeffery A. Reinbolt, and Terese L. Chmielewski. “Evaluation of a patient-specific cost function to predict the influence of foot path on the knee adduction torque during gait”. In: *Computer Methods in Biomechanics & Biomedical Engineering* 11.1 (Nov. 2007), pp. 63–71. DOI: 10.1080/10255840701552036. URL: https://doi.org/10.1080/10255840701552036.

[22] Kim L Bennell et al. “Higher dynamic medial knee load predicts greater cartilage loss over 12 months in medial knee osteoarthritis”. In: *Annals of the Rheumatic Diseases* 70.10 (July 2011), pp. 1770–1774. DOI: 10.1136/ard.2010.147082. URL: https://doi.org/10.1136/ard.2010.147082.

[23] Jonathan P. Walter et al. “Decreased knee adduction moment does not guarantee decreased medial contact force during gait”. In: *Journal of Orthopaedic Research®* 28.10 (Mar. 2010), pp. 1348–1354. DOI: 10.1002/jor.21142. URL: https://doi.org/10.1002/jor.21142.

[24] Andrew J. Meyer et al. “Are external knee load and EMG measures accurate indicators of internal knee contact forces during gait?” In: *Journal of Orthopaedic Research®* 31.6 (2013), pp. 921–929. DOI: 10.1002/jor.22304. URL: https://doi.org/10.1002/jor.22304.

[25] Mina Baniasad et al. “Knee adduction moment decomposition: Toward better clinical decision-making”. In: *Frontiers in Bioengineering and Biotechnology* 10 (Nov. 2022), p. 1017711. DOI: 10.3389/fbioe.2022.1017711. URL: https://doi.org/10.3389/fbioe.2022.1017711.

[26] Prasanna Sritharan, Yi-Chung Lin, and Marcus G. Pandy. “Muscles that do not cross the knee contribute to the knee adduction moment and tibiofemoral compartment loading during gait”. In: *Journal of Orthopaedic Research®* 30.10 (Mar. 2012), pp. 1586–1595. DOI: 10.1002/jor.22082. URL: https://doi.org/10.1002/jor.22082.

[27] Matt Prebble et al. “Simulated tibiofemoral joint reaction forces for three previously studied GAIT modifications in healthy controls”. In: *Journal of Biomechanical Engineering* 145.4 (Oct. 2022). DOI: 10.1115/1.4055885. URL: https://doi.org/10.1115/1.4055885.

[28] R.E. Richards et al. “Relationship between knee joint contact forces and external knee joint moments in patients with medial knee osteoarthritis: effects of gait modifications”. In: *Osteoarthritis and Cartilage* 26.9 (Apr. 2018), pp. 1203–1214. DOI: 10.1016/j.joca.2018.04.011. URL: https://doi.org/10.1016/j.joca.2018.04.011.

[29] Zachary F. Lerner et al. “How tibiofemoral alignment and contact locations affect predictions of medial and lateral tibiofemoral contact forces”. In: *Journal of Biomechanics* 48.4 (Jan. 2015), pp. 644–650. DOI: 10.1016/j.jbiomech.2014.12.049. URL: https://doi.org/10.1016/j.jbiomech.2014.12.049.

[30] Janelle M. Kaneda et al. “Can static optimization detect changes in peak medial knee contact forces induced by gait modifications?” In: *Journal of Biomechanics* 152 (Mar. 2023), p. 111569. DOI: 10.1016/j.jbiomech.2023.111569. URL: https://doi.org/10.1016/j.jbiomech.2023.111569.

[31] S. Meireles et al. “Knee contact forces are not altered in early knee osteoarthritis”. In: *Gait Posture* 45 (Jan. 2016), pp. 115–120. DOI: 10.1016/j.gaitpost.2016.01.016. URL: https://doi.org/10.1016/j.gaitpost.2016.01.016.

[32] Shinya Ogaya et al. “Knee adduction moment and medial knee contact force during gait in older people”. In: *Gait Posture* 40.3 (May 2014), pp. 341–345. DOI: 10.1016/j.gaitpost.2014.04.205. URL: https://doi.org/10.1016/j.gaitpost.2014.04.205.

[33] Samaneh Gholami et al. “Gait modification with subject-specific foot progression angle in people with moderate knee osteoarthritis: Investigation of knee adduction moment and muscle activity”. In: *The Knee* 35 (Mar. 2022), pp. 124–132. DOI: 10.1016/j.knee.2022.03.001. URL: https://doi.org/10.1016/j.knee.2022.03.001.

[34] Erasmus MC. *WalkWell Gait ReTrainer*. URL: https://www.erasmusmc.nl/en/research/projects/walkwell-gait-retrainer#c2e67d45-f788-434f-839b-859d56374f21.

[35] F. Leboeuf et al. “The conventional gait model, an open-source implementation that reproduces the past but prepares for the future”. In: *Gait & Posture* 69 (Mar. 2019), pp. 235–241. DOI: 10.1016/j.gaitpost.2019.04.015. URL: https://doi.org/10.1016/j.gaitpost.2019.04.015.

[36] Floor E. Heijs. “Foot Progression Angle Estimation: An Algorithm for Foot-based Single-Sensor Application”. Master’s thesis. Delft University of Technology, 2023. URL: https://resolver.tudelft.nl/uuid:c97101bc-de1c-4eea-b6f0-1cc35a30192c.

[37] Scott L. Delp et al. “OpenSiM: Open-Source software to create and analyze dynamic simulations of movement”. In: *IEEE Transactions on Biomedical Engineering* 54.11 (Oct. 2007), pp. 1940–1950. DOI: 10.1109/tbme.2007.901024. URL: https://doi.org/10.1109/tbme.2007.901024.

[38] Ajay Seth et al. “OpenSim: Simulating musculoskeletal dynamics and neuromuscular control to study human and animal movement”. In: *PLoS Computational Biology* 14.7 (July 2018), e1006223. DOI: 10.1371/journal.pcbi.1006223. URL: https://doi.org/10.1371/journal.pcbi.1006223.

[39] Bruno L. S. Bedo et al. “A custom musculoskeletal model for estimation of medial and lateral tibiofemoral contact forces during tasks with high knee and hip flexions”. In: *Computer Methods in Biomechanics & Biomedical Engineering* 23.10 (May 2020), pp. 658–663. DOI: 10.1080/10255842.2020.1757662. URL: https://doi.org/10.1080/10255842.2020.1757662.

[40] *Getting Started with Inverse Kinematics - OpenSim Documentation - OpenSim*. URL: https://opensimconfluence.atlassian.net/wiki/spaces/OpenSim/pages/53090032/Getting+Started+with+Inverse+Kinematics.

[41] *Getting Started with Static Optimization - OpenSim Documentation - OpenSim*. URL: https://opensimconfluence.atlassian.net/wiki/spaces/OpenSim/pages/53089624/Getting+Started+with+Static+Optimization.

[42] *Joint Reactions Analysis - OpenSiM Documentation - OpenSim*. URL: https://opensimconfluence.atlassian.net/wiki/spaces/OpenSim/pages/53089600/Joint+Reactions+Analysis.

[43] H. Marouane and A. Shirazi-Adl. “Sensitivity of medial-lateral load sharing to changes in adduction moments or angles in an asymptomatic knee joint model during gait”. In: *Gait Posture*

70 (Feb. 2019), pp. 39–47. DOI: 10.1016/j.gaitpost.2019.02.006. URL: https://doi.org/10.1016/j.gaitpost.2019.02.006.

[44] Todd C. Pataky. "Generalized n-dimensional biomechanical field analysis using statistical parametric mapping". In: *Journal of Biomechanics* 43.10 (May 2010), pp. 1976–1982. DOI: 10.1016/j.jbiomech.2010.03.008. URL: https://doi.org/10.1016/j.jbiomech.2010.03.008.

[45] John J Wilson and Masaru Furukawa. "Evaluation of the patient with hip pain". In: *American family physician* 89.1 (2014), pp. 27–34.

[46] Raziyeh Baghi et al. "Determining the individual relationship between the step width and peak knee adduction moment during stepping in medial knee osteoarthritis". In: *Clinical Biomechanics* 128 (July 2025), p. 106619. DOI: 10.1016/j.clinbiomech.2025.106619. URL: https://doi.org/10.1016/j.clinbiomech.2025.106619.

[47] Matthew S. DeMers, Saikat Pal, and Scott L. Delp. "Changes in tibiofemoral forces due to variations in muscle activity during walking". In: *Journal of Orthopaedic Research®* 32.6 (Feb. 2014), pp. 769–776. DOI: 10.1002/jor.22601. URL: https://doi.org/10.1002/jor.22601.

[48] Scott D Uhlrich et al. "Personalization improves the biomechanical efficacy of foot progression angle modifications in individuals with medial knee osteoarthritis". In: *Journal of Biomechanics* 144 (Nov. 2022), p. 111312. DOI: 10.1016/j.jbiomech.2022.111312. URL: https://doi.org/10.1016/j.jbiomech.2022.111312.

[49] D. Kumar, K.T. Manal, and K.S. Rudolph. "Knee joint loading during gait in healthy controls and individuals with knee osteoarthritis". In: *Osteoarthritis and Cartilage* 21.2 (Nov. 2012), pp. 298–305. DOI: 10.1016/j.joca.2012.11.008. URL: https://doi.org/10.1016/j.joca.2012.11.008.

[50] Susana Meireles et al. "Medial knee loading is altered in subjects with early osteoarthritis during gait but not during step-up-and-over task". In: *PLoS ONE* 12.11 (Nov. 2017), e0187583. DOI: 10.1371/journal.pone.0187583. URL: https://doi.org/10.1371/journal.pone.0187583.

[51] Jana Holder et al. "A systematic review of the associations between inverse dynamics and musculoskeletal modeling to investigate joint loading in a clinical environment". In: *Frontiers in Bioengineering and Biotechnology* 8 (Dec. 2020), p. 603907. DOI: 10.3389/fbioe.2020.603907. URL: https://doi.org/10.3389/fbioe.2020.603907.

[52] Ali H. Alnahdi, Joseph A. Zeni, and Lynn Snyder-Mackler. "Muscle impairments in patients with knee osteoarthritis". In: *Sports Health A Multidisciplinary Approach* 4.4 (June 2012), pp. 284–292. DOI: 10.1177/1941738112445726. URL: https://doi.org/10.1177/1941738112445726.

[53] Tibor Hortobágyi et al. "Altered hamstring-quadriceps muscle balance in patients with knee osteoarthritis". In: *Clinical Biomechanics* 20.1 (Oct. 2004), pp. 97–104. DOI: 10.1016/j.clinbiomech.2004.08.004. URL: https://doi.org/10.1016/j.clinbiomech.2004.08.004.

[54] Cheryl Hubley-Kozey, Kevin Deluzio, and Michael Dunbar. "Muscle co-activation patterns during walking in those with severe knee osteoarthritis". In: *Clinical Biomechanics* 23.1 (Dec. 2007), pp. 71–80. DOI: 10.1016/j.clinbiomech.2007.08.019. URL: https://doi.org/10.1016/j.clinbiomech.2007.08.019.

[55] Joseph A. Zeni, Katherine Rudolph, and Jill S. Higginson. "Alterations in quadriceps and hamstrings coordination in persons with medial compartment knee osteoarthritis". In: *Journal of Electromyography and Kinesiology* 20.1 (Feb. 2009), pp. 148–154. DOI: 10.1016/j.jelekin.2008.12.003. URL: https://doi.org/10.1016/j.jelekin.2008.12.003.

[56] M. Adouni and A. Shirazi-Adl. "Partitioning of knee joint internal forces in gait is dictated by the knee adduction angle and not by the knee adduction moment". In: *Journal of Biomechanics* 47.7 (Mar. 2014), pp. 1696–1703. DOI: 10.1016/j.jbiomech.2014.02.028. URL: https://doi.org/10.1016/j.jbiomech.2014.02.028.

[57] *SIMTK: OpenSim JAM: Project Home*. URL: https://simtk.org/projects/opensim-jam.

[58] Peter B. Shull et al. "Muscle force modification strategies are not consistent for gait retraining to reduce the knee adduction moment in individuals with knee osteoarthritis". In: *Journal of Biomechanics* 48.12 (July 2015), pp. 3163–3169. DOI: 10.1016/j.jbiomech.2015.07.006. URL: https://doi.org/10.1016/j.jbiomech.2015.07.006.

[59] S.P. Messier et al. “Influences of alignment and obesity on knee joint loading in osteoarthritic gait”. In: *Osteoarthritis and Cartilage* 22.7 (May 2014), pp. 912–917. DOI: 10.1016/j.joca.2014.05.013. URL: https://doi.org/10.1016/j.joca.2014.05.013.

[60] *GBD results*. URL: https://vizhub.healthdata.org/gbd-results/.