Simulated tibiofemoral joint reaction forces for three previously studied gait modifications in healthy controls

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ABSTRACT

Gait modifications, such as lateral trunk lean (LTL), medial knee thrust (MKT) and toe-in gait (TIG), are frequently investigated interventions used to slow the progression of knee osteoarthritis. The Lerner knee model was developed to estimate the tibiofemoral joint reaction forces (JRF) in the medial and lateral compartments during gait. These models may be useful for estimating the effects on the JRF in the knee as a result of gait modifications. We hypothesized that all gait modifications would decrease the JRF compared to normal gait. Twenty healthy individuals volunteered for this study (26.7 ± 4.7 years, 1.75 ± 0.1 m, 73.4 ± 12.4 kg). Ten trials were collected for normal gait as well as for the three gait modifications: LTL, MKT, and TIG. The data was used to estimate the JRF in the first and second peaks for the medial and lateral compartments of the knee via OpenSim using the Lerner knee model. No significant difference from baseline was found for the first peak in the medial compartment. There was a decrease in JRF in the medial compartment during the loading phase of gait for TIG (6.6%) and LTL (4.9%) and an increasing JRF for MKT (2.6%) but none were statistically significant. A significant increase from baseline was found for TIG (5.8%) in the medial second peak. We found a large variation in individual responses to gait interventions which may help explain the lack of statistically significant results. Possible factors influencing these wide range of responses to gait modifications include static alignment and the impacts of variation in muscle coordination strategies used, by participants, to implement gait modifications.

Keywords: *Gait modification, knee osteoarthritis, joint reaction forces, OpenSim*

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INTRODUCTION

Osteoarthritis (OA) of the knee is a major cause of disability and affects more than 19% of the adult population over the age of 45 in the United States [1, 2]. Excessive joint reaction forces (JRF) have been implicated in the development and progression of knee OA [3, 4]. Gait modifications are a promising non-invasive intervention used to reduce JRFs in the knee as evidence suggests that they may slow the progression of the disease [5-7]. A number of gait modifications have been identified that may help reduce the JRF in the medial compartment of the knee. Three common modifications investigated include lateral trunk lean (LTL) [6-14], medial knee thrust (MKT) [7-9, 11, 15-19], and toe-in gait (TIG) [7-9, 11, 17, 20]. A systematic review of gait interventions [7] found that when paired with real-time biofeedback (RTB), the three interventions chosen for this study show the greatest potential to reduce the knee adduction moment (KAM) in patients with knee OA. Real-time biofeedback can help improve the adoption of gait interventions and allow for the magnitude of modifications to be tailored to subject specific responses.

Many studies investigating gait modifications rely on surrogate measures, such as KAM and the knee flexion moment (KFM), to assess knee loads [17, 21-29]. In studies investigating knee OA it has been demonstrated that there is an association between KAM is and disease progression [30-33] while the contribution of KFM is less clear. One study found that both KAM and KFM equally contributed to estimates of total knee joint moments but that as the disease progressed KAM influenced increased while KFM influence decreased [30]. While some studies have found a connection between KFM and knee OA, especially in the first peak [20, 32], others have failed to show a strong connection [31, 34].

The LTL modification has been shown to reduce KAM when the trunk angle is modified by a sufficient amount [6, 9, 35, 36] but it has also been reported that the modification may lead to discomfort in the spine and ipsilateral knee and hip joints [6]. The MKT modification with real-time visual feedback has also been shown to reduce KAM in healthy subjects [9, 15, 36, 37]. A subject specific simulation study on a patient with grade 2 medial knee OA suggested that MKT could reduce both the first and second knee adduction torque peaks [16]. After gait retraining the patient was able to closely reproduce the knee adduction torque reductions, calculated by the simulation study, while walking in a laboratory setting [16]. The TIG modification using RTB has been shown to also reduce peak KAM in some studies [17, 35], however, other research has not demonstrated a significant decrease in KAM when using TIG with visual RTB [9, 38]. A study in healthy controls using TIG without RTB showed a statistically significant reduction in first peak KAM but found no reduction in JRF as estimated by a musculoskeletal model [39]. While some studies have found that KAM may provide a reasonable indicator for the JRF at the first peak of stance, the relationship between KAM and the joint contact force is not as strong for the second peak of stance [20, 34]. Additionally, research in children and adolescent's has indicated that KAM may not be a good predictor of knee joint contact force independent of leg alignment [40].

With the advancement in computing power computational models are becoming a common approach to directly estimate JRFs in the knee during gait [41-48]. While it is impractical to measure JRFs in-vivo [49], computational models are capable of estimating internal forces during functional movements (i.e. walking, running, crouch gait) [45, 46, 50-54]. OpenSim is an open-source software application for modeling, simulating, and analyzing movement [47]. It provides a flexible and robust tool that can be used by researchers to simulate how altered movement patterns can affect internal joint loading [41-47, 55-64]. Lerner et al. (2015) developed an OpenSim model (Lerner model) that allows for the direct estimation of the medial and lateral JRF in the knee.

Previous research has identified muscle forces as a major determinant of simulated compressive tibiofemoral contact forces thus variations in muscle activity greatly influence the accuracy of knee JRF predictions [50, 65]. The weighted static optimization objective function minimizes the sum of squared muscle activations while incorporating individual muscle weighting values as described in detail in previous research [52]. A common finding in past studies was that a weighted static optimization approach provides improved results over the default OpenSim static optimization (*SO*) [44, 45, 52, 66] by allowing researchers to reduce the force output in muscles that are causing increased error in JRF estimates. Many of these studies use in vivo data from instrumented knee implants to determine the weights for the objective function by minimizing the difference between the experimental and in vivo data [44, 45]. This is an infeasible approach to take when trying to estimate the effects of gait interventions in healthy or pathological subjects where no in vivo data is available. In order to implement the Lerner model an alternative approach to using in vivo data is needed for determining appropriate weights for muscles that exhibit larger than expected activations and forces as calculated by *SO*. Ensuring a proper weight for the model is important to reduce the overestimation of JRF in the model.

The purpose of this study was to examine the effect of three commonly studied gait interventions using RTB on the simulated JRF in a cohort of healthy participants. We hypothesized that all three interventions would reduce the first peak joint reaction force compared to baseline values. A second purpose of the study was to evaluate a visual inspection method for identifying muscles and weights for the weighted *SO* approach.

METHODS

Participants

Twenty healthy individuals were recruited from the university community to participate in the study. Their dominant limb was determined by identifying their preferred leg in a kicking task [67]. Eligibility criteria included no reported knee, hip, or back pain that required treatment within the prior six months and no previous lower limb or back surgery. Exclusion criteria included any cognitive impairment that would impact motor learning and existing neurological or musculoskeletal impairments that would affect gait. All participants provided written informed consent prior to participation and the study was approved by the George Mason University Institution Review Board. Participant demographics are presented in Table 1.

Instrumentation

Prior to data collection, retroreflective markers were attached to the lower extremities and trunk (53 markers) as shown below in Fig. 1. One cluster was located on the lower back and four clusters were placed bilaterally on the thigh and shank segments. Twelve tracking markers were placed on various anatomical locations and ten markers were placed on the feet. Ten calibration markers were used for static and dynamic calibration trials. For the static calibration trial participants stood motionless on a single force plate with their feet parallel to the anterior–posterior axis of the laboratory. Visual3D software (C-Motion, Germantown MD, USA) used data from the static trial to generate a kinematic model for each participant which included the trunk, pelvis, thigh, shank, and foot segments. For the dynamic calibration trial participants complete three clockwise rotations of the pelvis [68]. The data from the dynamic calibration was used to estimate hip joint center for the model. Calibration markers were removed before data collection. During data collection marker trajectories were tracked using eight high-speed motion analysis cameras (Vicon, Oxford, England) sampling at 200 Hz. Ground reaction force (GRF) was collected using four floor embedded force plates sampling at 1000 Hz (Bertec, Columbus, OH). The force plates were aligned in a single 2.4 m long row.

Data Collection

Baseline trials

Participants walked at a self-selected speed along a 6-meter laboratory walkway. Timing gates (Brower Timing Systems, Draper, UT, USA) were positioned at the start and end of four in-line force plates (2.4 meters long) and were used to measure the average walking speed per trial. For a trial to be valid, one full contact with a force plate by the dominant limb was required. Participants completed 10 valid baseline trials.

Gait modification trials

Gait modification parameters were individualized for each participant using their mean and standard deviation (*sd*) from baseline trials. Ranges for each gait modification were created so that gait parameters fell within a range of 1–3 *sd* greater (TIG and LTL) or lesser (MKT) than baseline for the first five trials and 3–5 *sd* greater or lesser than baseline for the second five trials. The 1–3 *sd* range was considered a small modification while the 3–5 *sd* range was considered a large modification. In total six target ranges were calculated for each participant: small and large LTL, small and large MKT, and small and large TIG.

Standardized verbal instructions, as described in previous research [69], were provided before implementing each modification. Participants were allowed to complete as many practice trials as desired to become comfortable with each modification and additional verbal feedback was provided during practice trials as needed. Gait modification trials were completed in a fixed order: LTL, MKT, and TIG. Successful trials required at least one clean foot strike on the force plate, with the dominant limb, and an average gait speed ± 5% relative to baseline average speed. Unsuccessful trials were not counted towards the 10 required for each modification.

For each of the 3 gait modification strategies studied, participants performed ten trials using RTB. The visual feedback was delivered using Visual3D via a line graph projected on a wall in front of the lab walkway as shown in Fig. 2. The graph indicated the angle of the current gait modification parameter during the stance phase of gait and was updated during each step of the dominant limb. A range representing the lower and upper limits of the gait modification parameter (1–3 or 3–5 *sd*) was displayed on the graph (i.e. the green band in Fig. 2). Participants modified their gait according to provided cues so that the line representing the gait parameter fell within the calculated range. If during a trial a participant fell outside the provided range, they were instructed to adjust their gait on the subsequent trial.

Musculoskeletal Simulation of Walking

Recorded data were first imported to Visual3D (C-Motion, Germantown MD, USA) for pre-processing as described in previous research [9]. OpenSim compatible format files were then exported from Visual3D. Prior to export, Visual3D runs inverse kinematics on the data and provides kinematic and kinetic .mot files for each trial. The exported files were used to create three dimensional simulations for the stance phase of gait using OpenSim 3.2. To simulate the muscle forces required to reproduce the measured kinematics and kinetics *SO* was run on the data using OpenSim 3.2. Prior to *SO*, the gait2392 model was scaled to each subject's height and weight. In addition to the default *SO* cost function for minimizing the sum of the muscle activations squared [47] each trial was also iteratively run through a weighted *SO* function based on previously described methods [44, 45, 52]. OpenSim 3.2 was used for this process because the weighted *SO* plug-in was built to be compatible with this version of the software and has not yet been updated to work with the latest version of OpenSim [70].

In order to identify muscles and corresponding weights the results of *SO* were visually inspected to identify any muscles identified in previous research that contributed to increase knee load estimates (e.g. quadriceps's, hamstrings, calves) and that had a force that was 2 to 3 times greater than estimated forces from other lower extremity muscle groups. A weight of 2 was initially applied to that muscle group and a weighted *SO* was re-run and the results visually inspected to determine the effect of the weight on the muscle force outputs. The weight was increased until the weighted *SO* output for the identified muscle fell within a comparable range to the other lower extremity muscles.

If there were multiple muscles with extreme force estimates, then muscles were weighted and evaluated in a set order for all gait intervention trials. The order was to apply a weight to the gastrocnemius (GAS) muscles followed by weights applied to the GAS and the vastus lateralis (GAS-VL) muscle, then to the GAS, VL, and vastus intermedius/vastus medialis (GAS-VI-VL-VM) muscles, and finally to the GAS, VL, VI, VM, and the rectus femoris (GAS-VI-VL-VM-RF) muscles. The muscle weight values started at two and were systematically adjusted until the muscle force output values from the weighted *SO* did not show any large spikes in the *SO* output.

After the default *SO* and the weighted *SO* was completed, the knee JRFs for the medial and lateral compartment were computed using the OpenSim JointReaction analyses on the scaled Lerner model using OpenSim 3.3 which is capable of resolving the JRFs in the knee into medial and lateral components [44].

Statistical Analysis

Descriptive statistics were reported while a within-group repeated measures analysis of variance (RM ANOVA) was used to compare JRF of participants' dominant limb across the four different gait conditions. A RM ANOVA was used with both the medial and lateral JRF for the first and second peaks during the stance phase of gait.

Prior to running the RM ANOVA, data was checked for outliers and normality. Next, Mauchly's test of sphericity was assessed. If the data failed the assumption of sphericity, a Greenhouse-Geisser correction was used. If results were significant for the RM ANOVA pairwise comparisons were calculated. Statistical analyses were performed using the ggstatsplot [71] package in R version 4.1.0 (R Foundation, Vienna, Austria, https://www.R-project.org) with an alpha level set at 0.05 *a priori*.

RESULTS

Mean JRF by gait conditions for the first and second peaks in both the medial and lateral knee compartment are shown below in Table 2. Post hoc analysis of the data indicated that the averages for JRF in both the small and large conditions did not differ significantly from each other, therefore, the results were combined into a single average across the three interventions for statistical analysis. Therefore, the statistical analysis consisted of 10 trials each for the baseline, LTL, MKT, and TIG.. Previous analysis indicated that subjects had a difficult time getting the modification to accurately fall within the prescribed bandwidth but were generally able to meet the lower bound of the prescribed modification [9].

The main effects of the RM ANOVA for the vertical JRFs are presented in Fig. 3 through 66. For the first peak JRF in the medial compartment there was no significant difference between conditions ($F(1.7, 32.3)=1.70$, $p=.20$). For the second peak in the medial compartment there was a statistically significant difference between conditions ($F(1.8, 34.4)=4.71$, $p=0.02$). Pairwise comparisons indicated that the TIG condition had a smaller JRF compared to baseline ($p=0.04$).

For the first peak JRF in the lateral compartment there was a statistically significant difference between conditions (F(1.8, 34.7)=10.56, p=0.0004). Pairwise comparisons indicated an increased JRF for MKT compared to baseline (p=0.01). For the second peak in the lateral compartment there was a statistically significant difference between conditions (F(3.0, 57.0)=3.81, p=0.01). However, pairwise comparisons indicated no significant difference between any condition and baseline.

Mean JRF (normalized by body weight) during the stance phase of gait is shown for the four gait conditions in both the medial and lateral compartments in Fig. 7.

DISCUSSION

This study compared the effects of three gait modifications on the simulated JRFs in the medial compartments of the knee in healthy participants. The primary purpose was to determine if LTL, MKT, and TIG reduced the JRFs in the medial compartment of the knee in healthy individuals. Our hypothesis was not supported by the data, which showed no statistically significant difference between baseline and any of the gait interventions in the medial compartment during the loading phase (e.g. first peak) of gait.

For individuals at risk for, or diagnosed with, medial compartment knee *OA*, reducing the JRF in the first peak is generally thought to be of high importance and gait interventions are commonly found to reduce either KAM or JRF in the first peak. While there was decreasing JRF in our study, the results were not statistically significant. The LTL modification has been shown to reduce KAM in healthy and pathological populations [9, 12, 72, 73]. One study did find that a healthy participant, when given verbal queues, could increase their JRF by using LTL type modification [74]. While this is counter to what would be expected we had a similar finding in our unpublished study that used data from a participant with an instrumented knee implant performing LTL.

In contrast to the LTL, the MKT modification showed an increasing JRF but it was also not significant. The MKT has been shown to decrease both KAM and JRF in previous studies [9, 15, 16, 75, 76]. One issue related to MKT is that of the three studied gait modifications we found MKT was the most difficult for participants to adopt, and therefore, the lack of results in our study could be due to inconsistent implementation of the MKT intervention [9]. If some participants were not able to correctly implement MKT it could have led to spurious data that obscured the results of the entire group. Overall, our results for MKT contradict previous research [72, 76, 77] with one possible explanation being that the outcome of gait interventions can be subject-specific [9, 78] and may be influenced by parameters such as anatomical alignment [79, 80], body mass index [81, 82], and individual gait biomechanics [78]. From a modeling perspective calculated JRF values are influenced by muscle activation and force estimates [74] so increased muscle activity, used by the model to match participant kinematics, during stance phase in MKT may influence the magnitude of the JRF in the first peak.

During TIG participants exhibited decreasing JRF but the results were also not statistically significant. Previous research on KAM in TIG has been inconclusive [78, 83-86] so these results were not unsurprising. However, a recent study that used musculoskeletal modeling with TIG found no change in JRF even with a decrease in KAM so our results are consistent with their findings [39].

While the second peak in the medial compartment may be less important for subjects with knee OA [20, 34], our data indicated a statistically significant difference between conditions for the propulsion phase of gait in the medial compartment (p=.02). The post-hoc pairwise comparisons indicated that TIG in the second peak produced a greater JRF then baseline (p=.04). Previous research on TIG has found inconclusive results on the effect of TIG on KAM so this isn't necessarily unexpected [11]. As shown in previous research the Lerner knee model may overestimate the JRF in the second peak of the medial compartment in normal gait [44] and unpublished data from our lab found larger error in the second peak when using MKT or LTL gait. While the weighted *SO* can reduce the error greatly [44, 87] the errors can still be 20% or large, as compared to the less than 10% error in the medial compartment for MKT and LTL gait.

One of the main goals of these gait interventions is to reduce the JRF in the medial compartment, however, a consequence may be that the load is transferred to another compartment. While our data did not find statistically significant reductions in JRFs in the medial compartment, there was a statistically significant difference from baseline in the lateral compartment for both the first (p=.004) and second (p=.01) peaks. Post hoc tests indicated that the JRF during MKT increased from baseline (p=.01) in the first peak but did not increase from baseline in the second peak. One factor to consider for this data is that in an unpublished validation study with the model it was found to overestimate JRF in the lateral compartment to a greater extent than in the first peak of the medial compartment, especially when applied to MKT and LTL interventions.

Therefore, the results in the lateral compartment may be skewed due to limitations of the model.

Data from previous research [87] suggests that this modeling approach can provide robust results for simulated JRFs in the medial compartment during the loading phase. Therefore, one possible explanation for lack of results is the intervention does not have a large effect on JRFs in the knee during the first peak. Another possibility is that there could be a large variation in individual responses to gait interventions that are averaged out in the aggregated data. Fig. 8 shows a summary of the responses to each intervention as a percentage increase or decrease from their baseline JRF values by study participant. Three participants had large (i.e. >30%) reductions in their baseline values while five had large increases. In addition, some individuals showed an increase from baseline for some gait modifications but a decrease for others. Previous research has reported a similar finding when evaluating KAM in all three modifications [9]. There has also been research that showed individual variation in the response to toe-in gait [78] and toe-out gait [85]. It is possible that no single variable contributes to an individual's response but that it is a combination of several variables. For example, a participant's response may be affected by the level of strength in their leg muscles interacting with static knee alignment, BMI, and/or other variables.

While research has demonstrated that gait modifications with RTB may help reduce the JRF in the medial compartment of the knee one limitation to this approach is that the majority of feedback is focused on joint kinematics, such as joint angles. Muscle contraction across a joint also contributes to JRF and recent research has shown that providing RTB of muscle activation patterns allows individuals to alter their JRF by changing the activation patterns of their muscles during gait [88]. This recent study found that participants were able to reduce knee JRF, during late stance, in a normal walking gait by 12% (±12) by shifting their muscle activation patterns from the gastrocnemius to the soleus muscle. It is possible that in gait modification studies participants are using different muscle coordination strategies to meet the RTB targets during modification. For example, someone trying to use MKT may unintentionally increase their knee JRF as a result of the particular muscle contraction strategies they use to meet the kinematics of the movement pattern. In addition to factors like static alignment these differences in movement coordination strategies may be contributing to the results shown in Fig 8 and offer a possible additional consideration for why some participants show decreasing JRF while others show increasing JRF. An alternative approach that may be worth investigating is providing RTB of both kinematic and muscle activation patterns, to subjects performing gait modifications, in order to gain a better understanding of each element's relative contribution to JRF.

There are several limitations to this study. First, while the results of the visual inspection method for identifying muscles and corresponding weights for the weighted *SO* was successful, it lacked objective metrics for decision criteria. An improved approach to the process would be to take a statistical measure, such as the mean of all lower extremity muscles forces, during the second half of stance, and for any muscle with an excessive force estimate increase the weight until the muscle force estimate is within 1 or 2 standard deviations of that median value. However, more work should be done to further evaluate this process for identifying muscle weighting and identifying the optimal appropriate range could be developed experimentally. Another limitation to this analysis was a lack of imaging data for participants. This did not allow us to accurately determine static knee alignment and contact locations and reduced the benefit of the Lerner model. The Lerner model has modifiable parameters which lets you adjust the knee alignment in the frontal plane as well as the contact locations between the tibia and femur. While we did attempt to estimate the values from Visual3D, the estimate of knee alignment calculation from the software differs from the approach described by Lerner et al [44, 89] in their validation study and we were not confident enough in the values to use them for this study. Lastly, it is also possible that the sample size may not have been large enough to provide the power to detect differences in our data.

CONCLUSION

This study did not find the hypothesized decrease in simulated JRF in the medial compartment of the knee for TIG, LTL, and MKT. Lack of results may be as a result of individual variation in response to gait modifications. Factors involved in this variation may include static alignment of the lower extremity or muscle coordination patterns used to meet gait modification targets. It is possible that gait interventions cannot be uniformly applied to all participants with the expectation that they will all respond with reductions in JRF. Individual factors may lead some participants to have decreased JRF as a result of a modification while others may experience increased JRF. It is also possible that there may be variation within an individual in how they respond to different gait modifications with some leading to reduced JRF and others increasing JRF. Future work should be done to develop a greater understanding of how different factors contribute to individual responses in JRFs as a result of gait modification.

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NOMENCLATURE

BMI	Body mass index
F	Female
GAS	Gastrocnemius
GAS-VL	Gastrocnemius and vastus lateralis
GAS-VI-VL-VM	Gastrocnemius, vastus intermedius vastus lateralis, and vastus medialis
GAS-VI-VL-VM-RF	Gastrocnemius, vastus intermedius vastus lateralis, vastus medialis, and rectus femoris
JRF	Joint reaction force
KAM	Knee adduction moment
KFM	Knee flexion moment

kg	Kilogram
L	Left
LTL	Lateral trunk lean
M	Male
m	Meter
MKT	Medial knee thrust
N	Number
OA	Osteoarthritis
R	Right
RM ANOVA	Repeated measures analysis of variance
sd	Standard deviation
SO	Static optimization
TIG	Toe-in gait

REFERENCES

[1] Cross M., Smith E., Hoy D., Nolte S., Ackerman I., Fransen M., Bridgett L., Williams S., Guillemin F., Hill C. L., Laslett L. L., Jones G., Cicuttini F., Osborne R., Vos T., Buchbinder R., Woolf A., March L., 2014, "The global burden of hip and knee osteoarthritis: estimates from the global burden of disease 2010 study," Ann Rheum Dis, 73(7): 1323-30. https://doi.org/10.1136/annrheumdis-2013-204763

[2] Lawrence R. C., Felson D. T., Helmick C. G., Arnold L. M., Choi H., Deyo R. A., Gabriel S., Hirsch R., Hochberg M. C., Hunder G. G., Jordan J. M., Katz J. N., Kremers H. M., Wolfe F., National Arthritis Data W., 2008, "Estimates of the prevalence of arthritis and other rheumatic conditions in the United States. Part II," Arthritis Rheum, 58(1): 26-35. https://doi.org/10.1002/art.23176

[3] Ratzlaff C. R., Koehoorn M., Cibere J., Kopec J. A., 2012, "Is lifelong knee joint force from work, home, and sport related to knee osteoarthritis?," Int J Rheumatol, 2012: 584193. https://doi.org/10.1155/2012/584193

[4] Sharma L., Hurwitz D. E., Thonar E. J., Sum J. A., Lenz M. E., Dunlop D. D., Schnitzer T. J., Kirwan-Mellis G., Andriacchi T. P., 1998, "Knee adduction moment, serum hyaluronan level, and disease severity in medial tibiofemoral osteoarthritis," Arthritis Rheum, 41(7): 1233-40. https://doi.org/10.1002/1529-0131(199807)41:7<1233::AID-ART14>3.0.CO;2-L

[5] Ferrigno C., Stoller I. S., Shakoor N., Thorp L. E., Wimmer M. A., 2016, "The feasibility of using augmented auditory feedback from a pressure detecting insole to reduce the knee adduction moment: A proof of concept study," J Biomech Eng, 138(2): 021014. https://doi.org/10.1115/1.4032123

[6] Hunt M. A., Simic M., Hinman R. S., Bennell K. L., Wrigley T. V., 2011, "Feasibility of a gait retraining strategy for reducing knee joint loading: increased trunk lean guided by real-time biofeedback," J Biomech, 44(5): 943-47. https://doi.org/10.1016/j.jbiomech.2010.11.027

[7] Eddo O., Lindsey B., Caswell S. V., Cortes N., 2017, "Current evidence of gait modification with real-time biofeedback to alter kinetic, temporospatial, and function-related outcomes: A review," International Journal of Kinesiology & Sports Science, 5(3): 21-35. https://doi.org/10.7575/aiac.ijkss.v.5n.3p.35

[8] Fregly B. J., 2012, "Gait modification to treat knee osteoarthritis," HSS J, 8(1): 45-48. https://doi.org/10.1007/s11420-011-9229-9

[9] Lindsey B., Eddo O., Caswell S. V., Prebble M., Cortes N., 2020, "Reductions in peak knee abduction moment in three previously studied gait modification strategies," Knee, 27(1): 102-10. https://doi.org/10.1016/j.knee.2019.09.017

[10] Miller R. H., Esterson A. Y., Shim J. K., 2015, "Joint contact forces when minimizing the external knee adduction moment by gait modification: A computer simulation study," Knee, 22(6): 481-89. https://doi.org/10.1016/j.knee.2015.06.014

[11] Simic M., Hinman R. S., Wrigley T. V., Bennell K. L., Hunt M. A., 2011, "Gait modification strategies for altering medial knee joint load: a systematic review," Arthritis Care Res (Hoboken), 63(3): 405-26. https://doi.org/10.1002/acr.20380

[12] Simic M., Hunt M. A., Bennell K. L., Hinman R. S., Wrigley T. V., 2012, "Trunk lean gait modification and knee joint load in people with medial knee osteoarthritis: the effect of varying trunk lean angles," Arthritis Care Res (Hoboken), 64(10): 1545-53. https://doi.org/10.1002/acr.21724

[13] Hunt M. A., Birmingham T. B., Bryant D., Jones I., Giffin J. R., Jenkyn T. R., Vandervoort A. A., 2008, "Lateral trunk lean explains variation in dynamic knee joint load in patients with medial compartment knee osteoarthritis," Osteoarthritis Cartilage, 16(5): 591-99. https://doi.org/10.1016/j.joca.2007.10.017
[14] Mundermann A., Asay J. L., Mundermann L., Andriacchi T. P., 2008, "Implications of increased medio-lateral trunk sway for ambulatory mechanics," J Biomech, 41(1): 165-70. https://doi.org/10.1016/j.jbiomech.2007.07.001
[15] Ferrigno C., Wimmer M. A., Trombley R. M., Lundberg H. J., Shakoor N., Thorp L. E., 2016, "A reduction in the knee adduction moment with medial thrust gait is associated with a medial shift in center of plantar pressure," Med Eng Phys, 38(7): 615-21. https://doi.org/10.1016/j.medengphy.2016.03.008
[16] Fregly B. J., Reinbolt J. A., Rooney K. L., Mitchell K. H., Chmielewski T. L., 2007, "Design of patient-specific gait modifications for knee osteoarthritis rehabilitation," IEEE Trans Biomed Eng, 54(9): 1687-95. https://doi.org/10.1109/TBME.2007.907637
[17] Jackson B., Gordon K. E., Chang A. H., 2018, "Immediate and short-term effects of real-time knee adduction moment feedback on the peak and cumulative knee load during walking," J Orthop Res, 36(1): 397-404. https://doi.org/10.1002/jor.23659
[18] Kinney A. L., Besier T. F., Silder A., Delp S. L., D'Lima D. D., Fregly B. J., 2013, "Changes in in vivo knee contact forces through gait modification," J Orthop Res, 31(3): 434-40. https://doi.org/10.1002/jor.22240
[19] Walter J. P., D'Lima D. D., Colwell C. W., Jr., Fregly B. J., 2010, "Decreased knee adduction moment does not guarantee decreased medial contact force during gait," J Orthop Res, 28(10): 1348-54. https://doi.org/10.1002/jor.21142
[20] Richards R. E., Andersen M. S., Harlaar J., van den Noort J. C., 2018, "Relationship between knee joint contact forces and external knee joint moments in patients with medial knee osteoarthritis: effects of gait modifications," Osteoarthritis Cartilage, 26(9): 1203-14. https://doi.org/10.1016/j.joca.2018.04.011
[21] Tan H. H., Mentiplay B., Quek J. J., Tham A. C. W., Lim L. Z. X., Clark R. A., Woon E. L., Yeh T. T., Tan C. I. C., Pua Y. H., 2020, "Test-retest reliability and variability of knee adduction moment peak, impulse and loading rate during walking," Gait Posture, 80: 113-16. https://doi.org/10.1016/j.gaitpost.2020.05.029
[22] Wang S., Chan K. H. C., Lam R. H. M., Yuen D. N. S., Fan C. K. M., Chu T. T. C., Baur H., Cheung R. T. H., 2019, "Effects of foot progression angle adjustment on external knee adduction moment and knee adduction angular impulse during stair ascent and descent," Hum Mov Sci, 64: 213-20. https://doi.org/10.1016/j.humov.2019.02.004
[23] Tomoya T., Mutsuaki E., Takuma I., Yuta T., Masayoshi K., 2019, "A mathematical modelling study investigating the influence of knee joint flexion angle and extension moment on patellofemoral joint reaction force and stress," Knee, 26(6): 1323-29. https://doi.org/10.1016/j.knee.2019.10.010
[24] Nie Y., Wang H., Xu B., Zhou Z., Shen B., Pei F., 2019, "The relationship between knee adduction moment and knee osteoarthritis symptoms according to static alignment and pelvic drop," Biomed Res Int, 2019: 7603249. https://doi.org/10.1155/2019/7603249

[25] Konrath J. M., Karatsidis A., Schepers H. M., Bellusci G., de Zee M., Andersen M. S., 2019, "Estimation of the knee adduction moment and joint contact force during daily living activities using inertial motion capture," Sensors (Basel), 19(7): 1681. https://doi.org/10.3390/s19071681
[26] Luc-Harkey B. A., Franz J. R., Blackburn J. T., Padua D. A., Hackney A. C., Pietrosimone B., 2018, "Real-time biofeedback can increase and decrease vertical ground reaction force, knee flexion excursion, and knee extension moment during walking in individuals with anterior cruciate ligament reconstruction," J Biomech, 76: 94-102. https://doi.org/10.1016/j.jbiomech.2018.05.043
[27] Telfer S., Lange M. J., Sudduth A. S. M., 2017, "Factors influencing knee adduction moment measurement: A systematic review and meta-regression analysis," Gait Posture, 58: 333-39. https://doi.org/10.1016/j.gaitpost.2017.08.025
[28] Teng H. L., MacLeod T. D., Link T. M., Majumdar S., Souza R. B., 2015, "Higher knee flexion moment during the second half of the stance phase of gait is associated with the progression of osteoarthritis of the patellofemoral joint on magnetic resonance imaging," J Orthop Sports Phys Ther, 45(9): 656-64. https://doi.org/10.2519/jospt.2015.5859
[29] Creaby M. W., 2015, "It's not all about the knee adduction moment: the role of the knee flexion moment in medial knee joint loading," Osteoarthritis Cartilage, 23(7): 1038-40. https://doi.org/10.1016/j.joca.2015.03.032
[30] Asay J. L., Erhart-Hledik J. C., Andriacchi T. P., 2018, "Changes in the total knee joint moment in patients with medial compartment knee osteoarthritis over 5 years," J Orthop Res, 36(9): 2373-79. https://doi.org/10.1002/jor.23908
[31] Chang A. H., Moisio K. C., Chmiel J. S., Eckstein F., Guermazi A., Prasad P. V., Zhang Y., Almagor O., Belisle L., Hayes K., Sharma L., 2015, "External knee adduction and flexion moments during gait and medial tibiofemoral disease progression in knee osteoarthritis," Osteoarthritis Cartilage, 23(7): 1099-106. https://doi.org/10.1016/j.joca.2015.02.005
[32] Chehab E. F., Favre J., Erhart-Hledik J. C., Andriacchi T. P., 2014, "Baseline knee adduction and flexion moments during walking are both associated with 5 year cartilage changes in patients with medial knee osteoarthritis," Osteoarthritis Cartilage, 22(11): 1833-39. https://doi.org/10.1016/j.joca.2014.08.009
[33] Erhart-Hledik J. C., Chehab E. F., Asay J. L., Favre J., Chu C. R., Andriacchi T. P., 2021, "Longitudinal changes in tibial and femoral cartilage thickness are associated with baseline ambulatory kinetics and cartilage oligomeric matrix protein (COMP) measures in an asymptomatic aging population," Osteoarthritis Cartilage, 29(5): 687-96. https://doi.org/10.1016/j.joca.2021.02.006
[34] Holder J., Trinler U., Meurer A., Stief F., 2020, "A systematic review of the associations between inverse dynamics and musculoskeletal modeling to investigate joint loading in a clinical environment," Front Bioeng Biotechnol, 8: 603907. https://doi.org/10.3389/fbioe.2020.603907
[35] Shull P. B., Lurie K. L., Cutkosky M. R., Besier T. F., 2011, "Training multi-parameter gaits to reduce the knee adduction moment with data-driven models and haptic feedback," J Biomech, 44(8): 1605-09. https://doi.org/10.1016/j.jbiomech.2011.03.016

[36] Simic M., Hinman R. S., Wrigley T. V., Bennell K. L., Hunt M. A., 2011, "Gait modification strategies for altering medial knee joint load: a systematic review," Arthritis Care Res (Hoboken), 63(3): 405-26. https://doi.org/10.1002/acr.20380

[37] Barrios J. A., Crossley K. M., Davis I. S., 2010, "Gait retraining to reduce the knee adduction moment through real-time visual feedback of dynamic knee alignment," J Biomech, 43(11): 2208-13. https://doi.org/10.1016/j.jbiomech.2010.03.040

[38] Richards R., van den Noort J. C., van der Esch M., Booij M. J., Harlaar J., 2018, "Gait retraining using real-time feedback in patients with medial knee osteoarthritis: Feasibility and effects of a six-week gait training program," Knee, 25(5): 814-24. https://doi.org/10.1016/j.knee.2018.05.014

[39] Cui W., Wang C., Chen W., Guo Y., Jia Y., Du W., Wang C., 2019, "Effects of Toe-Out and Toe-In Gaits on Lower-Extremity Kinematics, Dynamics, and Electromyography," Applied Sciences, 9(23): 5245.

[40] Holder J., Drongelen S., Meurer A., Stief F. Statistical comparison of contact forces and moments in the knee joint during walking in participants with and without valgus malalignment. ESMAC: Gait & Posture, 2020:155-56.

[41] Eskinazi I., Fregly B. J., 2016, "An open-source toolbox for surrogate modeling of joint contact mechanics," IEEE Trans Biomed Eng, 63(2): 269-77. https://doi.org/10.1109/TBME.2015.2455510

[42] Xu H., Bloswick D., Merryweather A., 2015, "An improved OpenSim gait model with multiple degrees of freedom knee joint and knee ligaments," Comput Methods Biomech Biomed Engin, 18(11): 1217-24. https://doi.org/10.1080/10255842.2014.889689

[43] Martelli S., Valente G., Viceconti M., Taddei F., 2015, "Sensitivity of a subject-specific musculoskeletal model to the uncertainties on the joint axes location," Comput Methods Biomech Biomed Engin, 18(14): 1555-63. https://doi.org/10.1080/10255842.2014.930134

[44] Lerner Z. F., DeMers M. S., Delp S. L., Browning R. C., 2015, "How tibiofemoral alignment and contact locations affect predictions of medial and lateral tibiofemoral contact forces," J Biomech, 48(4): 644-50. https://doi.org/10.1016/j.jbiomech.2014.12.049

[45] Knarr B. A., Higginson J. S., 2015, "Practical approach to subject-specific estimation of knee joint contact force," J Biomech, 48(11): 2897-902. https://doi.org/10.1016/j.jbiomech.2015.04.020

[46] Gerus P., Sartori M., Besier T. F., Fregly B. J., Delp S. L., Banks S. A., Pandy M. G., D'Lima D. D., Lloyd D. G., 2013, "Subject-specific knee joint geometry improves predictions of medial tibiofemoral contact forces," J Biomech, 46(16): 2778-86. https://doi.org/10.1016/j.jbiomech.2013.09.005

[47] Delp S. L., Anderson F. C., Arnold A. S., Loan P., Habib A., John C. T., Guendelman E., Thelen D. G., 2007, "OpenSim: open-source software to create and analyze dynamic simulations of movement," IEEE Trans Biomed Eng, 54(11): 1940-50. https://doi.org/10.1109/TBME.2007.901024

[48] Otten E., 2003, "Inverse and forward dynamics: models of multi-body systems," Philos Trans R Soc Lond B Biol Sci, 358(1437): 1493-500. https://doi.org/10.1098/rstb.2003.1354

[49] Hart D. A., Martin C. R., Scott M., Shrive N. G., 2021, "The instrumented sheep knee to elucidate insights into osteoarthritis development and progression: A sensitive and reproducible platform for integrated research efforts," Clin Biomech (Bristol, Avon), 87: 105404. https://doi.org/10.1016/j.clinbiomech.2021.105404

[50] Demers M. S., Pal S., Delp S. L., 2014, "Changes in tibiofemoral forces due to variations in muscle activity during walking," J Orthop Res, 32(6): 769-76. https://doi.org/10.1002/jor.22601

[51] Lerner Z. F., Haight D. J., DeMers M. S., Board W. J., Browning R. C., 2014, "The effects of walking speed on tibiofemoral loading estimated via musculoskeletal modeling," J Appl Biomech, 30(2): 197-205. https://doi.org/10.1123/jab.2012-0206

[52] Steele K. M., Demers M. S., Schwartz M. H., Delp S. L., 2012, "Compressive tibiofemoral force during crouch gait," Gait Posture, 35(4): 556-60. https://doi.org/10.1016/j.gaitpost.2011.11.023

[53] Seth A., Sherman M., Reinbolt J. A., Delp S. L., 2011, "OpenSim: a musculoskeletal modeling and simulation framework for in silico investigations and exchange," Procedia IUTAM, 2: 212-32. https://doi.org/10.1016/j.piutam.2011.04.021

[54] Gu W., Pandy M. G., 2020, "Direct validation of human knee-joint contact mechanics derived from subject-specific finite-element models of the tibiofemoral and patellofemoral joints," J Biomech Eng, 142(7): 071001. https://doi.org/10.1115/1.4045594

[55] Meireles S., De Groote F., Reeves N. D., Verschueren S., Maganaris C., Luyten F., Jonkers I., 2016, "Knee contact forces are not altered in early knee osteoarthritis," Gait Posture, 45: 115-20. https://doi.org/10.1016/j.gaitpost.2016.01.016

[56] Haight D. J., Lerner Z. F., Board W. J., Browning R. C., 2014, "A comparison of slow, uphill and fast, level walking on lower extremity biomechanics and tibiofemoral joint loading in obese and nonobese adults," J Orthop Res, 32(2): 324-30. https://doi.org/10.1002/jor.22497

[57] Richards C., Higginson J. S., 2010, "Knee contact force in subjects with symmetrical OA grades: differences between OA severities," J Biomech, 43(13): 2595-600. https://doi.org/10.1016/j.jbiomech.2010.05.006

[58] Winby C. R., Lloyd D. G., Besier T. F., Kirk T. B., 2009, "Muscle and external load contribution to knee joint contact loads during normal gait," J Biomech, 42(14): 2294-300. https://doi.org/10.1016/j.jbiomech.2009.06.019

[59] Uicker J. J., Sheth, P. N. *Matrix methods in the design analysis of multibody systems*. Charlottesville, VA: University of Virginia, 2007.

[60] Shelburne K. B., Torry M. R., Pandy M. G., 2006, "Contributions of muscles, ligaments, and the ground-reaction force to tibiofemoral joint loading during normal gait," J Orthop Res, 24(10): 1983-90. https://doi.org/10.1002/jor.20255

[61] Taylor W. R., Heller M. O., Bergmann G., Duda G. N., 2004, "Tibio-femoral loading during human gait and stair climbing," J Orthop Res, 22(3): 625-32. https://doi.org/10.1016/j.orthres.2003.09.003

[62] Hurwitz D. E., Sumner D. R., Andriacchi T. P., Sugar D. A., 1998, "Dynamic knee loads during gait predict proximal tibial bone distribution," J Biomech, 31(5): 423-30. https://doi.org/10.1016/S0021-9290(98)00028-1

[63] Schipplein O. D., Andriacchi T. P., 1991, "Interaction between active and passive knee stabilizers during level walking," J Orthop Res, 9(1): 113-19. https://doi.org/10.1002/jor.1100090114
[64] Morrison J. B., 1970, "The mechanics of the knee joint in relation to normal walking," Journal of Biomechanics, 3(1): 51-61. https://doi.org/10.1016/0021-9290(70)90050-3
[65] Herzog W., Longino D., Clark A., 2003, "The role of muscles in joint adaptation and degeneration," Langenbecks Arch Surg, 388(5): 305-15. https://doi.org/10.1007/s00423-003-0402-6
[66] Knarr B. A., Higginson J. S., Zeni J. A., 2016, "Change in knee contact force with simulated change in body weight," Comput Methods Biomech Biomed Engin, 19(3): 320-23. https://doi.org/10.1080/10255842.2015.1018193
[67] Cortes N., Quammen D., Lucci S., Greska E., Onate J., 2012, "A functional agility short-term fatigue protocol changes lower extremity mechanics," J Sports Sci, 30(8): 797-805. https://doi.org/10.1080/02640414.2012.671528
[68] Schwartz M. H., Rozumalski A., 2005, "A new method for estimating joint parameters from motion data," J Biomech, 38(1): 107-16. https://doi.org/10.1016/j.jbiomech.2004.03.009
[69] Eddo O. O., Lindsey B. W., Caswell S. V., Prebble M., Cortes N., 2019, "Unintended Changes in Contralateral Limb as a Result of Acute Gait Modification," J Appl Biomech, 36(1): 13-19. https://doi.org/10.1123/jab.2019-0031
[70] Steele K. M., Tresch M. C., Perreault E. J., 2015, "Consequences of biomechanically constrained tasks in the design and interpretation of synergy analyses," J Neurophysiol, 113(7): 2102-13. https://doi.org/10.1152/jn.00769.2013
[71] Patil I., 2021, "Visualizations with statistical details: The 'ggstatsplot' approach.," Journal of Open Source Software, 6(61): 3167. https://doi.org/10.21105/joss.03167
[72] Gerbrands T. A., Pisters M. F., Theeven P. J. R., Verschueren S., Vanwanseele B., 2017, "Lateral trunk lean and medializing the knee as gait strategies for knee osteoarthritis," Gait Posture, 51: 247-53. https://doi.org/10.1016/j.gaitpost.2016.11.014
[73] Tokuda K., Anan M., Takahashi M., Sawada T., Tanimoto K., Kito N., Shinkoda K., 2018, "Biomechanical mechanism of lateral trunk lean gait for knee osteoarthritis patients," J Biomech, 66: 10-17. https://doi.org/10.1016/j.jbiomech.2017.10.016
[74] Pizzolato C., Reggiani M., Saxby D. J., Ceseracciu E., Modenese L., Lloyd D. G., 2017, "Biofeedback for gait retraining based on real-time estimation of tibiofemoral joint contact forces," IEEE Trans Neural Syst Rehabil Eng, 25(9): 1612-21. https://doi.org/10.1109/TNSRE.2017.2683488
[75] Schache A. G., Fregly B. J., Crossley K. M., Hinman R. S., Pandy M. G., 2008, "The effect of gait modification on the external knee adduction moment is reference frame dependent," Clin Biomech (Bristol, Avon), 23(5): 601-08. https://doi.org/10.1016/j.clinbiomech.2007.12.008
[76] Fregly B. J., D'Lima D. D., Colwell C. W., Jr., 2009, "Effective gait patterns for offloading the medial compartment of the knee," J Orthop Res, 27(8): 1016-21. https://doi.org/10.1002/jor.20843
[77] Meyer A. J., D'Lima D. D., Besier T. F., Lloyd D. G., Colwell C. W., Jr., Fregly B. J., 2013, "Are external knee load and EMG measures accurate indicators of internal

knee contact forces during gait?," J Orthop Res, 31(6): 921-29. https://doi.org/10.1002/jor.22304

[78] Uhlrich S. D., Silder A., Beaupre G. S., Shull P. B., Delp S. L., 2018, "Subject-specific toe-in or toe-out gait modifications reduce the larger knee adduction moment peak more than a non-personalized approach," J Biomech, 66: 103-10. https://doi.org/10.1016/j.jbiomech.2017.11.003

[79] Hoch M. C., Weinhandl J. T., 2017, "Effect of valgus knee alignment on gait biomechanics in healthy women," J Electromyogr Kinesiol, 35: 17-23. https://doi.org/10.1016/j.jelekin.2017.05.003

[80] Clement J., Toliopoulos P., Hagemeister N., Desmeules F., Fuentes A., Vendittoli P. A., 2018, "Healthy 3D knee kinematics during gait: Differences between women and men, and correlation with x-ray alignment," Gait Posture, 64: 198-204. https://doi.org/10.1016/j.gaitpost.2018.06.024

[81] Sharma L., Lou C., Cahue S., Dunlop D. D., 2000, "The mechanism of the effect of obesity in knee osteoarthritis: the mediating role of malalignment," Arthritis Rheum, 43(3): 568-75. https://doi.org/10.1002/1529-0131(200003)43:3<568::AID-ANR13>3.0.CO;2-E

[82] Silva F. R., Muniz A. M. d. S., Cerqueira L. S., Nadal J., 2018, "Biomechanical alterations of gait on overweight subjects," Research on biomedical engineering, 34(4): 291-98. https://doi.org/10.1590/2446-4740.180017

[83] Lynn S. K., Kajaks T., Costigan P. A., 2008, "The effect of internal and external foot rotation on the adduction moment and lateral-medial shear force at the knee during gait," J Sci Med Sport, 11(5): 444-51. https://doi.org/10.1016/j.jsams.2007.03.004

[84] Bechard D. J., Birmingham T. B., Zecevic A. A., Jones I. C., Giffin J. R., Jenkyn T. R., 2012, "Toe-out, lateral trunk lean, and pelvic obliquity during prolonged walking in patients with medial compartment knee osteoarthritis and healthy controls," Arthritis Care Res (Hoboken), 64(4): 525-32. https://doi.org/10.1002/acr.21584

[85] Hunt M. A., Takacs J., 2014, "Effects of a 10-week toe-out gait modification intervention in people with medial knee osteoarthritis: a pilot, feasibility study," Osteoarthritis Cartilage, 22(7): 904-11. https://doi.org/10.1016/j.joca.2014.04.007

[86] Charlton J. M., Hatfield G. L., Guenette J. A., Hunt M. A., 2018, "Toe-in and toe-out walking require different lower limb neuromuscular patterns in people with knee osteoarthritis," J Biomech, 76: 112-18. https://doi.org/10.1016/j.jbiomech.2018.05.041

[87] Prebble M., Wei Q., Eddo O., Lindsey B., Caswell S. V., Cortes N., 2019, "Preliminary analysis: The effects of gait interventions on knee joint contact forces in healthy adults," Medicine & Science in Sports & Exercise, 51(6S): 703. https://doi.org/10.1249/01.mss.0000562592.89136.7b

[88] Uhlrich S. D., Jackson R. W., Seth A., Kolesar J. A., Delp S. L., 2022, "Muscle coordination retraining inspired by musculoskeletal simulations reduces knee contact force," Sci Rep, 12(1): 9842. https://doi.org/10.1038/s41598-022-13386-9

[89] Tsushima H., Morris M. E., McGinley J., 2003, "Test-retest reliability and inter-tester reliability of kinematic data from a three-dimensional gait analysis system," J Jpn Phys Ther Assoc, 6(1): 9-17. https://doi.org/10.1298/jjpta.6.9

Figure Captions List

Fig. 1 Marker placement for data collection. Individual tracking markers (22 markers) were attached to the manubrium, 7th cervical vertebrae, right scapula, 10th thoracic vertebrae, and bilaterally to the posterior and lateral calcaneus, 5th distal metatarsal, 1st proximal metatarsal, 2nd metatarsophalangeal joint, tibial tuberosity, lateral iliac spine, posterior superior iliac spine, and acromion. Three tracking markers were arranged to form a triangular cluster and were attached to the lumbar region. Four tracking clusters (18 markers) were placed on the lateral aspect of the thigh and shank. Ten calibration markers were attached bilaterally to the lateral and medial malleoli, lateral and medial knee joint lines, and greater trochanters[9]

Fig. 2 Example of visual feedback graph provided to participants that was projected onto the laboratory wall during each trial

Fig. 3 Repeated measures ANOVA for the mean joint reaction force (N) in the 1st peak in the medial compartment for baseline, lateral trunk lean, medial knee thrust, and toe-in gait

Fig. 4 Repeated measures ANOVA for the mean joint reaction force (N) in the 2nd peak in the medial compartment for baseline, lateral trunk lean, medial knee thrust, and toe-in gait

Fig. 5 Repeated measures ANOVA for the mean joint reaction force (N) in the 1st peak in the lateral compartment for baseline, lateral trunk lean, medial knee thrust, and toe-in gait

Fig. 6 Repeated measures ANOVA for the mean joint reaction force (N) in the 2nd peak in the lateral compartment for baseline, lateral trunk lean, medial knee thrust, and toe-in gait

Fig. 7 Mean joint reaction force (normalized by body weight) in the medial and lateral knee compartments for baseline, lateral trunk lean, medial knee thrust, and toe-in gait

Fig. 8 Percentage reduction in joint reaction force from baseline values, by individual participant, for toe-in gait, lateral trunk lean, and medial knee thrust

Table Caption List

Table 1 Participant characteristics

Table 2 Peak mean (±*sd*) joint reaction forces during gait for baseline, lateral trunk lean, medial knee thrust, and toe-in gait for the first and second peak in the medial compartment and the first and second peak in the lateral compartment

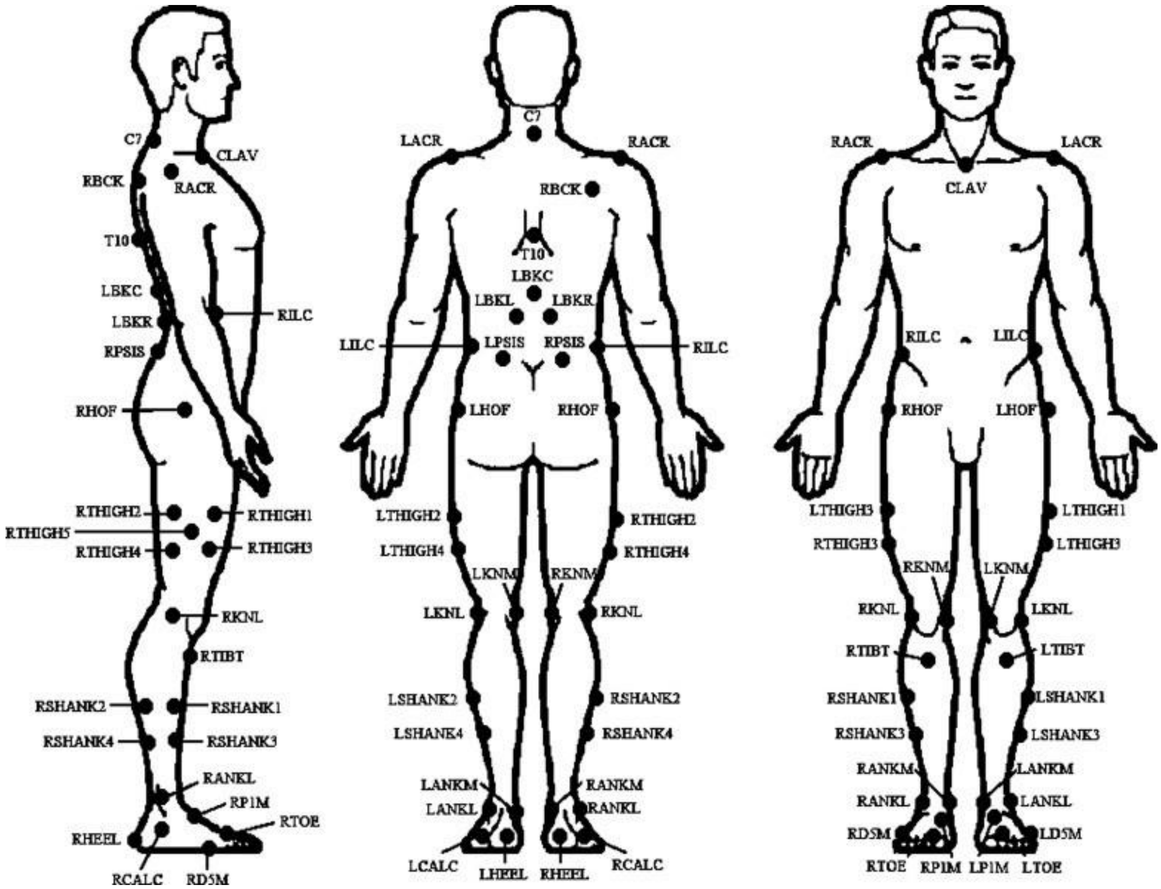


Fig. 1 Marker placement for data collection. Individual tracking markers (22 markers) were attached to the manubrium, 7th cervical vertebrae, right scapula, 10th thoracic vertebrae, and bilaterally to the posterior and lateral calcaneus, 5th distal metatarsal, 1st proximal metatarsal, 2nd metatarsophalangeal joint, tibial tuberosity, lateral iliac spine, posterior superior iliac spine, and acromion. Three tracking markers were arranged to form a triangular cluster and were attached to the lumbar region. Four tracking clusters (18 markers) were placed on the lateral aspect of the thigh and shank. Ten calibration markers were attached bilaterally to the lateral and medial malleoli, lateral and medial knee joint lines, and greater trochanters[9]

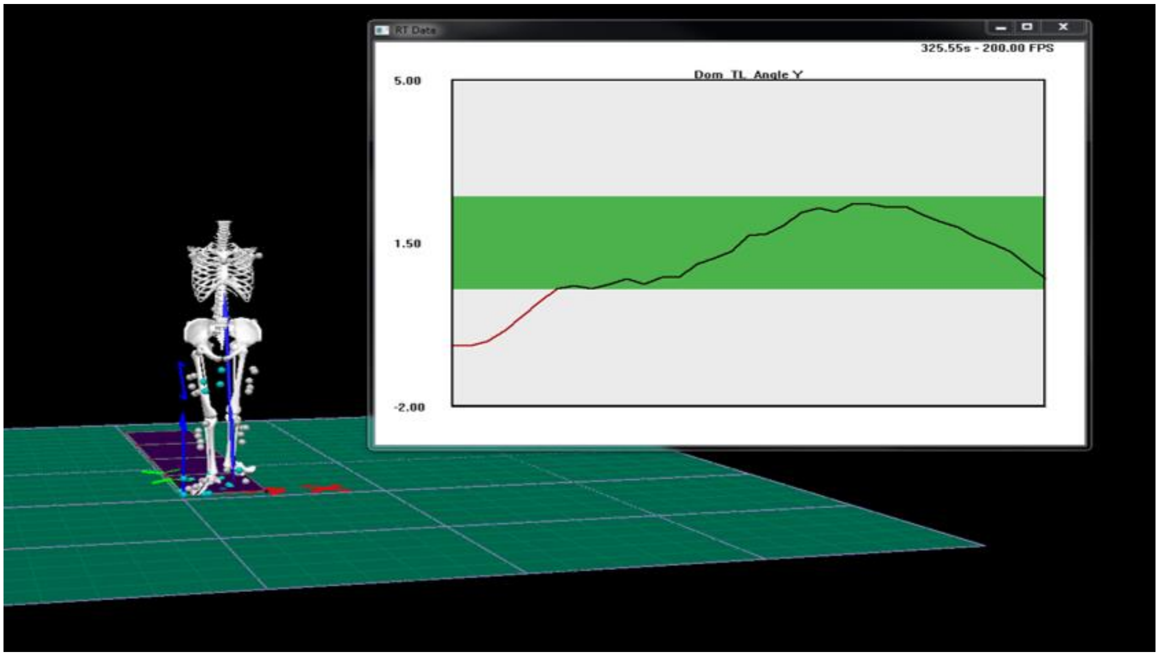
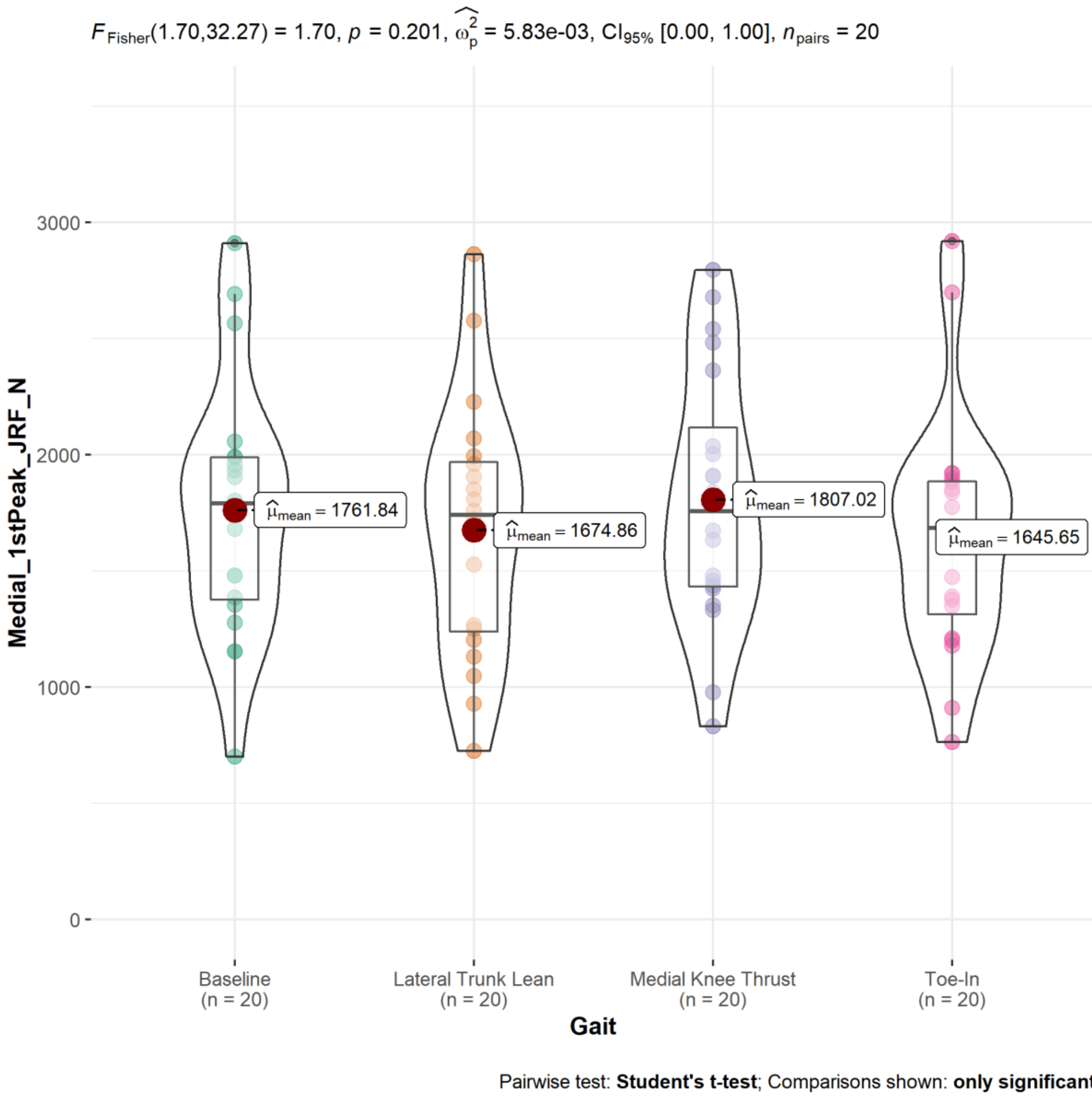


Fig. 2 Example of visual feedback graph provided to participants that was projected onto the laboratory wall during each trial



748
749
750 **Fig. 3 Repeated measures ANOVA for the mean joint reaction force (N) in the 1st**
751 **peak in the medial compartment for baseline, lateral trunk lean, medial knee thrust,**
752 **and toe-in gait**

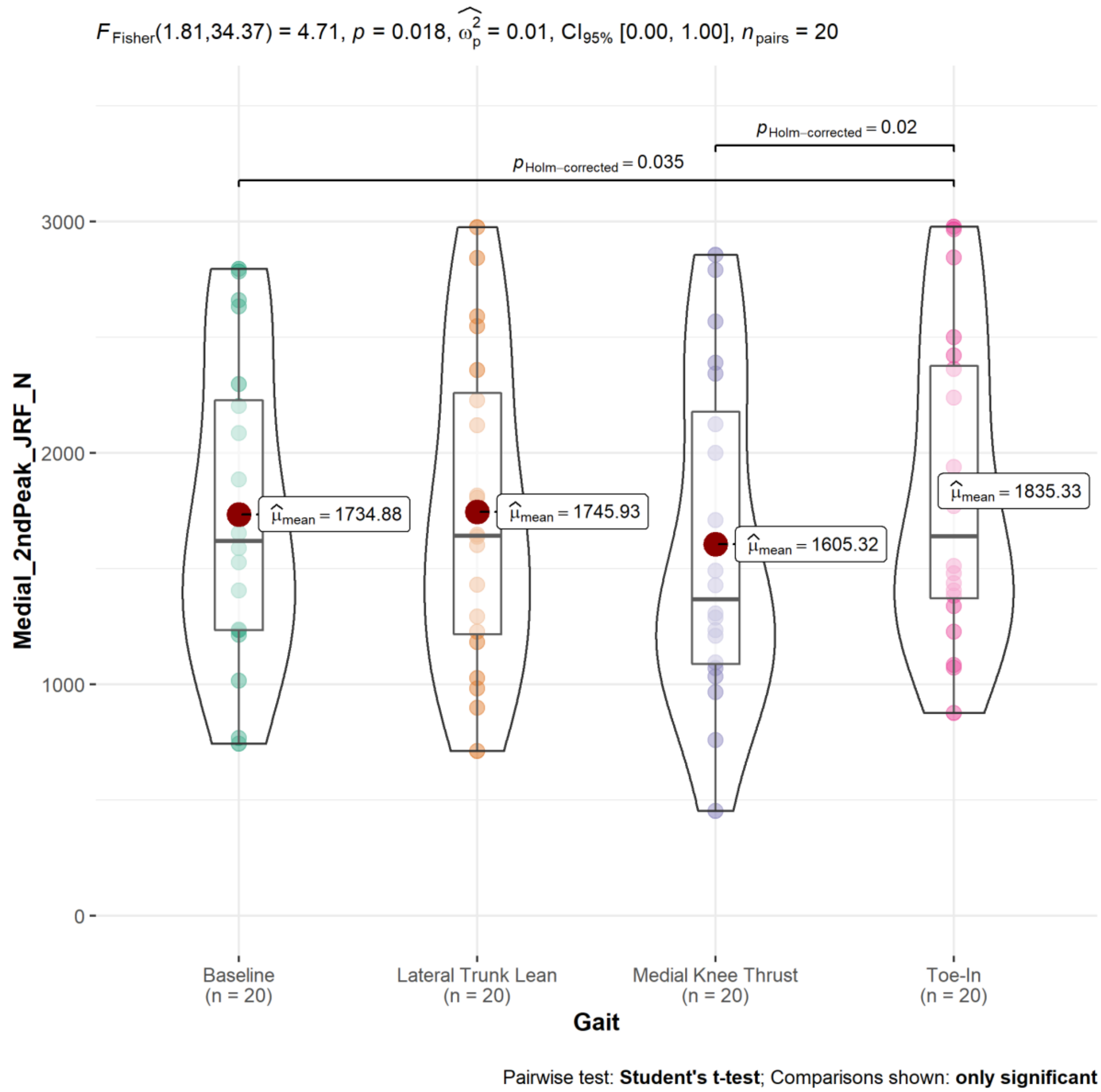


Fig. 4 Repeated measures ANOVA for the mean joint reaction force (N) in the 2nd peak in the medial compartment for baseline, lateral trunk lean, medial knee thrust, and toe-in gait

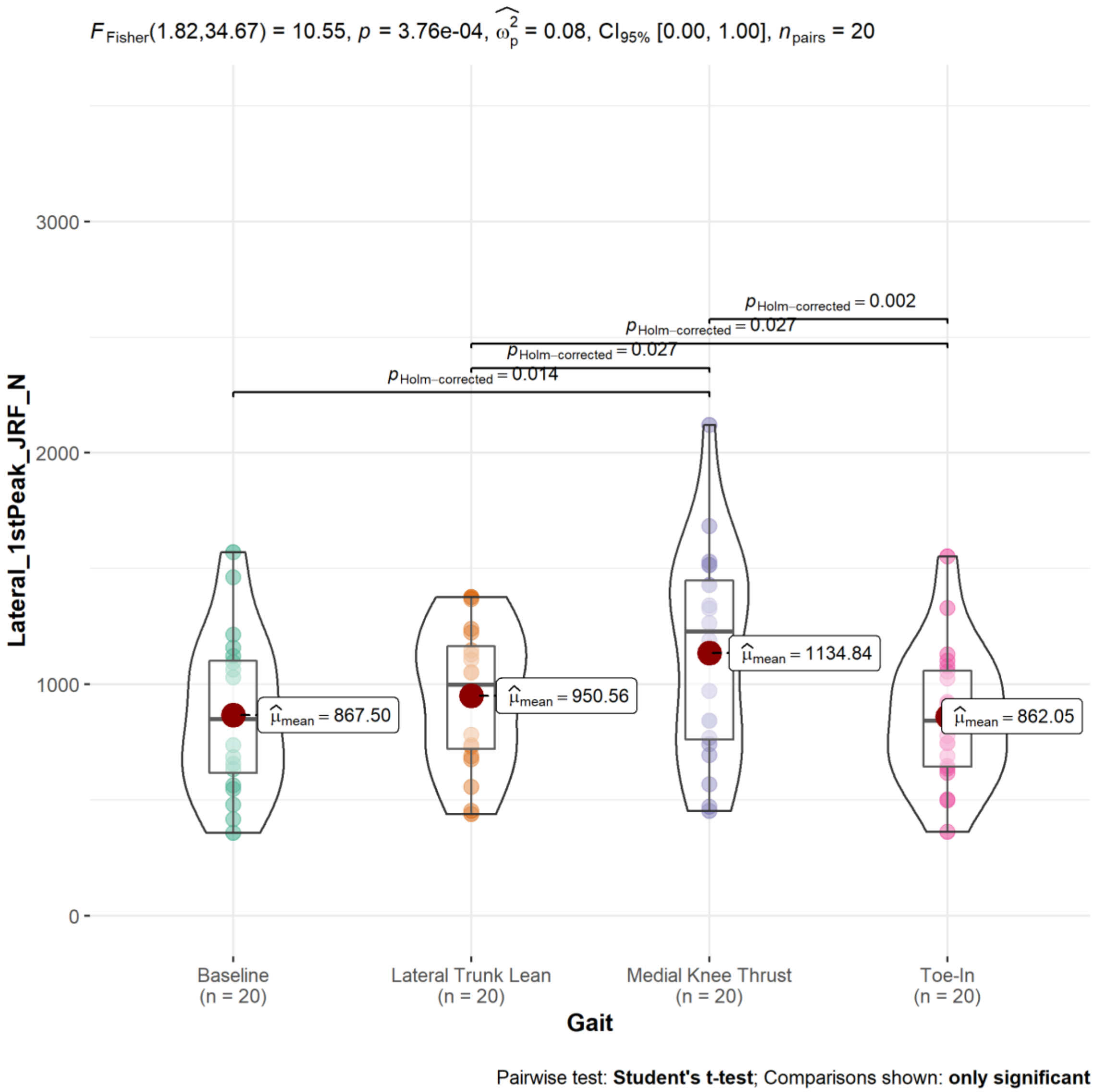


Fig. 5 Repeated measures ANOVA for the mean joint reaction force (N) in the 1st peak in the lateral compartment for baseline, lateral trunk lean, medial knee thrust, and toe-in gait

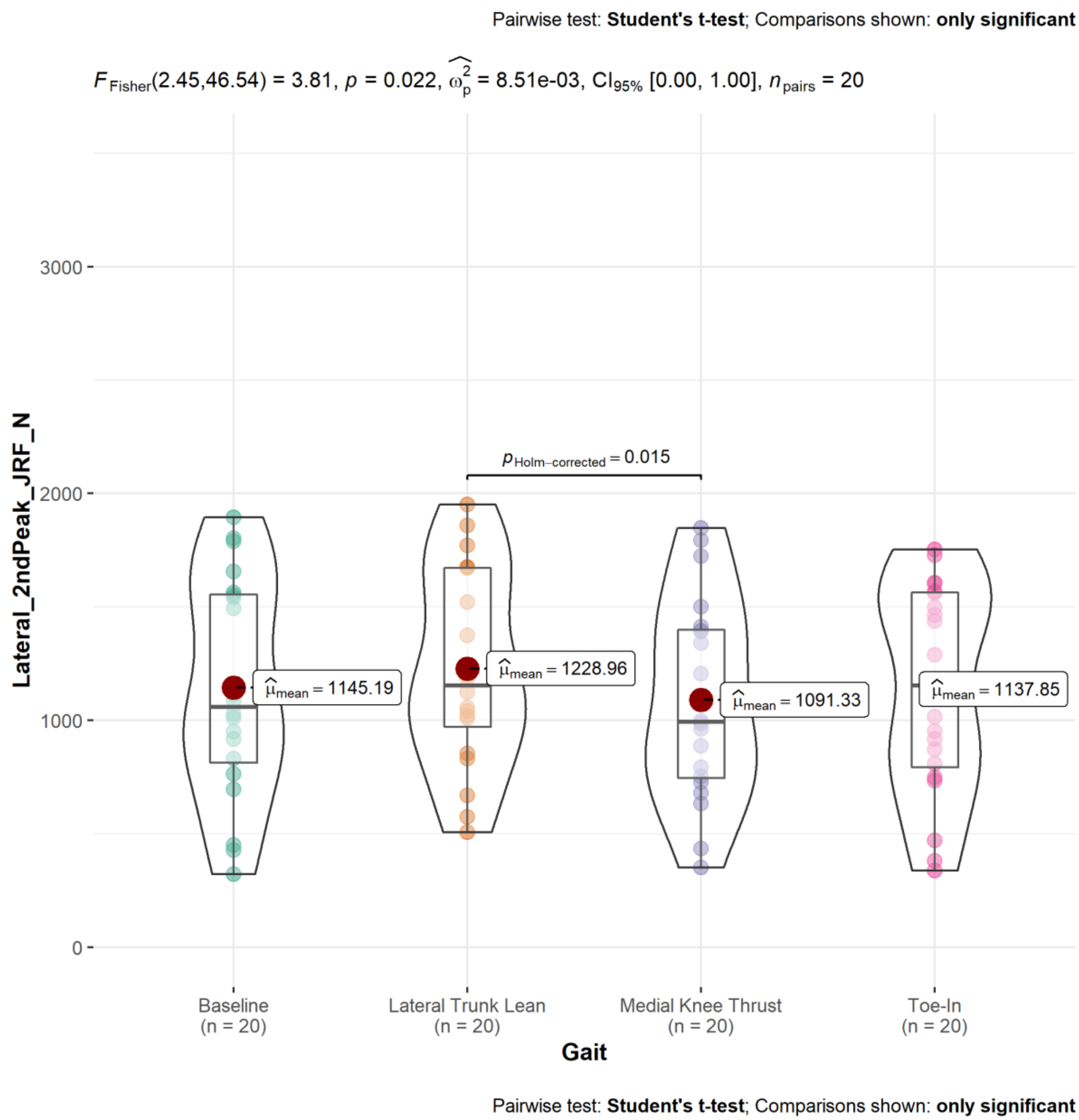


Fig. 6 Repeated measures ANOVA for the mean joint reaction force (N) in the 2nd peak in the lateral compartment for baseline, lateral trunk lean, medial knee thrust, and toe-in gait

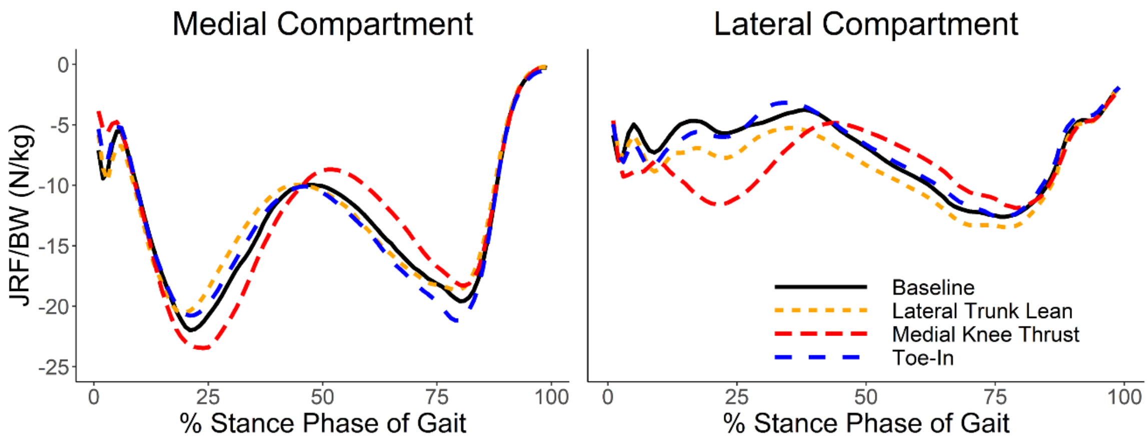


Fig. 7 Mean joint reaction force (normalized by body weight) in the medial and lateral knee compartments for baseline, lateral trunk lean, medial knee thrust, and toe-in gait

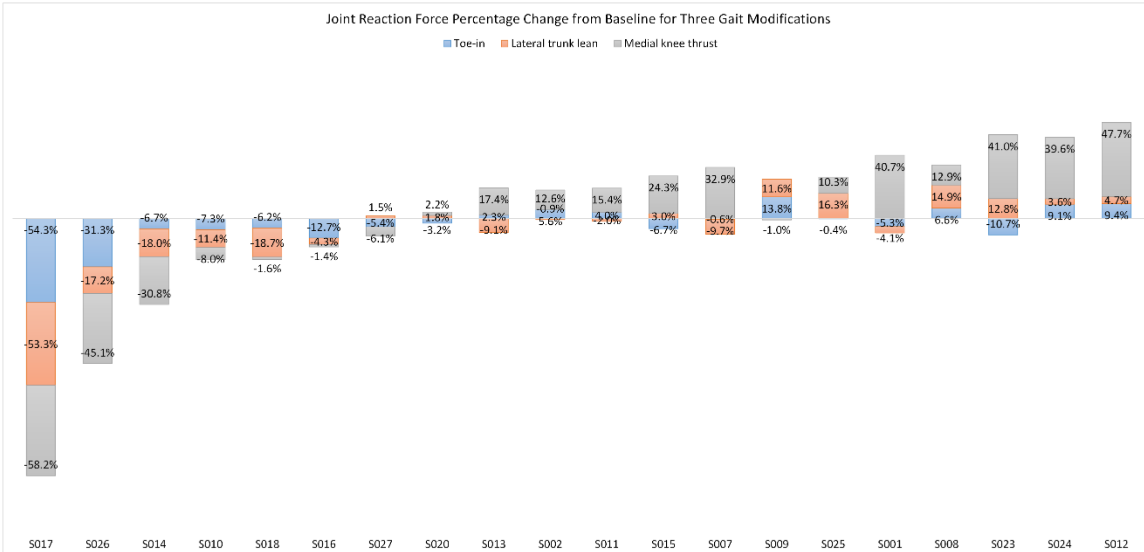


Fig. 8 Percentage reduction in joint reaction force from baseline values, by individual participant, for toe-in gait, lateral trunk lean, and medial knee thrust

Table 1 Participant characteristics

Characteristics	**Mean (*sd*)**
N	20
Gender (*M*/*F*)	12/8
Dominant Limb (*R*/*L*)	18/2
Age (years)	26.7 (4.7)
Height (*m*)	1.75 (0.1)
Mass (*kg*)	73.4 (12.4)
BMI	23.9 (3.0)

Table 2 Peak mean (±*sd*) joint reaction forces during gait for baseline, lateral trunk lean, medial knee thrust, and toe-in gait for the first and second peak in the medial compartment and the first and second peak in the lateral compartment

	Med_1stPeak_FY Mean (±*sd*)	Med_2ndPeak_FY Mean (±*sd*)	Lat_1stPeak_FY Mean (±*sd*)	Lat_2ndPeak_FY Mean (±*sd*)
Baseline	1761.84 (±166.40)	1734.88 (±170.65)	867.50 (±122.46)	1145.19 (±88.48)
LTL	1674.86 (±185.31)	1745.93 (±228.67)	950.56 (±164.65)	1228.96 (±143.37)
MKT	1807.02 (±249.57)	1605.32 (±245.75)	1134.84 (±183.07)	1091.33 (±161.62)
TIG	1645.65 (±159.61)	1835.33 (±182.29)	862.05 (±107.33)	1137.85 (±100.73)