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Gait changes associated with the reduced push-off from solid ankle foot orthoses

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Dissertation

GAIT CHANGES ASSOCIATED WITH THE REDUCED PUSH-OFF FROM SOLID ANKLE FOOT ORTHOSES

by

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ABSTRACT

Introduction

Ankle foot orthoses (AFOs) are used to improve walking in some lower extremity conditions but AFOs restrict ankle motion resulting in a trade-off in ankle and hip mechanics. While the use of AFOs have been well documented, there still remain gaps in the literature. The first study compared the differences in sagittal plane ankle and hip kinematics and kinetics across three conditions at two speeds in healthy individuals while the second study compared frontal plane kinetics at the hip and knee and vertical ground reaction forces between two conditions at two speeds in healthy individuals.

Methods

This was studied by collecting and analyzing three-dimensional joint kinematics and ground reaction forces from twelve healthy adults. Participants walked in three conditions (shod; i.e. athletic shoes only and two reduced push-off conditions using solid ankle foot orthoses (SAFOs); i.e. unilateral brace and bilateral brace conditions) and at two speeds (1.25m/s and 1.5m/s). In the first study, generalized linear models with general

estimating equations were used to compare ankle and hip angles, moments and power for the braced and unbraced sides separately in all three conditions. In the second study, frontal plane kinetics and vertical ground reaction forces in the unbraced limb in the unilateral brace condition were compared to the same side during shod walking using paired sample t-tests.

Results

From our first study we found that the reduced push-off from the use of SAFOs results in decreased peak plantarflexion angles and power generation at the ankle and increased peak flexion angles, and first and second peak power generation at the hip in the braced limbs in both unilateral ($p \leq 0.05$) and bilateral ($p \leq 0.05$) brace conditions at both speeds. On the unbraced side in the unilateral brace condition, there were decreased peak power generation at the ankle at 1.25m/s and increased peak extension moments, first and second peak power generation at the hip compared to the shod condition ($p<0.05$) at both speeds.

In the comparison between the unilateral and bilateral brace conditions, the changes in ankle and hip mechanics were similar to the changes between the shod condition and the bilateral brace condition on the unbraced side; in addition, participants also had higher peak extension moments in the unilateral brace condition compared to the bilateral brace condition ($p<0.05$). On the braced side, participants had lower peak plantarflexion moments at the ankle and lower peak flexion angles at the hip when walking with bilateral SAFOs, compared to walking with unilateral SAFOs ($p<0.05$).

In the second study, we found that peak internal knee and hip abduction moments were 3% and 4% higher, respectively, in the unbraced limb in the unilateral brace condition at 1.25m/s ($p \leq 0.041$) compared to the same side in the shod condition. Peak vertical ground reaction force was 3% higher in the unbraced limb in the unilateral brace condition at both speeds ($p=0.002$).

Conclusion

Findings indicate that walking with unilateral ankle foot orthoses presents an increased risk of developing secondary conditions.

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LIST OF ABBREVIATIONS

AAFOs	Articulated ankle foot orthoses
AFO	Ankle foot orthoses
ASIS	Anterior superior iliac spines
BMI	Body mass index
FRAFO	Floor reaction ankle foot orthoses
GEE	General estimating equations
GLM	Generalized linear models
GRAFO	Ground reaction ankle foot orthoses
H1	First peak hip power generation
H3	Second peak hip power generation
OA	Osteoarthritis
PLSAFO	Posterior leaf spring ankle foot orthoses
PSIS	Posterior superior iliac spines
SAFO	Solid Ankle foot orthoses
SSAFOs	Semi-solid ankle foot orthoses

CHAPTER ONE

1. Introduction

Healthy humans walk by generating, a large amount of the power at the ankle joint by the plantarflexors (i.e. soleus, lateral and medial gastrocnemius) during late stance often referred to as push-off. [1, 2, 3] In the United States, it is estimated that over 126 million individuals representing a half of the population live with musculoskeletal conditions. [4] Often times, these conditions (e.g. cerebral palsy, multiple sclerosis, Parkinson's disease, stroke) affect the foot, subtalar and/or ankle joint altering their normal function during walking. In individuals with lower extremity musculoskeletal conditions, the generation of push-off can be affected resulting in gait deviations which have been associated with increased metabolic costs [1] and increased risk of developing secondary conditions. [5, 6]

Ankle foot orthoses (AFOs) popularly referred to as leg braces are prescribed to improve walking in individuals with musculoskeletal pathologies that affect the normal function of the ankle, subtalar and/or knee joints. In lower extremity rehabilitation, the main goals of this intervention may be to control the range and rate of motion, correct a deformity, properly align, compensate for weakness or relieve pressure at the ankle, subtalar and/or knee joints. [7, 8] While these are usually the goals for rehabilitation, the restricted ankle motion from the use of AFOs are also known to reduce the amount of power generated by the ankle plantarflexors during push-off. [1, 9] Although there is extensive evidence documenting improved gait parameters with the use of AFOs in varied patient populations, [10, 11, 12, 13, 14] their effects have not been studied as much in healthy individuals. As AFOs reduce push-off, an insight into the biomechanical changes from walking with AFOs without the confounding effects of pathology is needed to provide a better understanding of the gait deviations observed in individuals with lower extremity musculoskeletal impairments who ambulate with AFOs.

1.1 Types of Ankle Foot Orthoses (AFOs)

There are several types of AFOs. In clinical practice, an orthotist selects an AFO to meet an individual's mobility demands based on their specific impairment. [7] The types of AFOs that a clinician may choose from include conventional metal systems, thermoplastic systems, carbon fiber systems, prefabricated systems and hybrid systems.

1.1.1 Thermoplastic Ankle Foot Orthoses

Across North America, thermoplastic AFOs are popularly prescribed and dispensed in clinics due to their light weight and high acceptance in patient populations. Types of thermoplastic AFOs include ground or floor reaction AFOs (GRAFOs or FRAFOs), posterior leaf spring AFOs (PLSAFOs), the solid AFOs (SAFOs), semi-solid AFOs (SSAFOs) and articulated AFOs (AAFOs).

The studies in this dissertation reduced push-off using solid ankle foot orthoses (SAFOs). SAFOs were used because they have been documented to produce large reductions in push-off generation [9] which provided an ideal scenario for studying the effects of reduced push-off on gait. At the ankle, SAFOs immobilize the joint holding it in a neutral position throughout the gait cycle. SAFOs control motion in all three planes (i.e. plantarflexion and dorsiflexion motions in the sagittal plane, abduction and adduction motions in the frontal plane and inversion and eversion in the transverse plane).

1.2 Reduced Push-off

Bipedal walking in humans is powered mainly at the ankle and hip joints. In healthy gait, a large burst of positive power is generated at the ankle joint during plantarflexion at the end of the stance phase. [1, 15] Research has shown that generating a majority of the power needed to walk at the ankle joint is a more efficient way to walk. [1, 2, 16, 17]

The amount of push-off generated is affected in some pathological gait and while AFOs are often prescribed to improve gait, these devices also have the tendency to reduce the amount of push generated. [1, 2, 17] This dissertation studied the effects of reduced push-off on lower extremity mechanics in unilateral and bilateral brace conditions using solid ankle foot orthoses.

1.3 Limitations of Previous Studies

Previous studies have shown that ankle power generation and plantarflexion moments decrease significantly with the use of AFOs. The observed decrease in ankle kinetics have been attributed to the restriction of the ankle joint and the needed plantar flexor strength to deform the AFO. [18] Few studies have investigated push-off using a brace often prescribed in clinics and no study has compared gait mechanics between unilateral and bilateral reduced push-off conditions in healthy individuals. Furthermore, this dissertation will include the first study that reports on frontal plane knee and hip kinetics and vertical ground reaction forces with the use of unilateral SAFOs: variables that have been linked with osteoarthritis.

1.4 Aims

My long-term goal is to better understand the biomechanical factors and mechanisms that result in gait deviations in individuals that use prosthetic or orthotic devices. As reduced push-off is a universal attribute in this population, establishing its effect on healthy gait will provide valuable information for achieving my long-term goal. The aim of this dissertation was to study the biomechanical changes associated with the reduced push-off from the use of solid ankle foot orthoses in healthy individuals. Generally, we hypothesized that a reduction in the amount of push-off at the ankle will lead to compensatory mechanics at the other lower extremity joints.

Study 1: Compare the walking adaptations between walking with athletic shoes (shod), walking with unilateral reduced push-off and walking with bilateral reduced push-off. This was done by comparing ankle and hip angles, moments and power in the braced and unbraced sides in the 3 conditions.

Hypotheses: We hypothesized that compared to shod walking the use of AFOs would result in decreased peak ankle angles, moments and power and increased peak hip angle, moments and power.

Study 2: Determine the changes in frontal plane knee and hip moments and vertical ground reaction force associated with the reduced push-off from unilateral SAFOs.

Hypotheses: We hypothesized that compared to shod walking the use of a unilateral SAFO will lead to an increase in peak internal abduction moments at the knee and hip in the unbraced limb in the unilateral brace condition. We also hypothesized that the use of the SAFO would result in increased vertical ground reaction forces in the unbraced limb in the unilateral brace condition compared to the corresponding side in the shod condition.

CHAPTER TWO

STUDY 1: CHANGES IN ANKLE AND HIP MECHANICS ASSOCIATED WITH THE REDUCED PUSH-OFF FROM SOLID ANKLE FOOT ORTHOSES

2.1 Introduction

Bipedal walking can be modeled by applying an impulsive push to the trailing limb or by generating a torque at the hip. [2] Powering gait at the trailing limb is a more efficient way to walk and it is known to be four times less expensive energetically compared to generating a torque at the hip. [2] Healthy humans walk by producing a large amount of power at the ankle joint during late stance often referred to as push-off, and a small amount at the hip by the extensors in the leading limb. [1, 3] Lower limb interventions may alter the generation of push-off and in such cases, individuals compensate at the hip. [3, 9]

Ankle push-off contributes to the acceleration of the trailing limb and to the redirection of the center of mass. [19] Varying the amount of push-off has been studied for clinical applications. [3, 20] Increasing push-off is known to decrease hip kinetics; a trade-off which may be advantageous in some patient populations. [20] Specifically, walking with increased push-off decreases hip extension moments [3] and anteriorly directed forces [20] and has been suggested to improve anterior hip pain. [20] Conversely, in scenarios where push-off is reduced, individuals walk with increased hip moments and power. [1, 9] Walking with increased hip extension has been observed in individuals with anterior hip pain. Lower extremity orthotic (unilateral or bilateral) interventions reduce push-off and thus, may have negative implications on the hip joint

Previous studies on reduced push-off in healthy individuals have reported reduced peak ankle power generation and increased peak hip extension moments and first peak hip power generation with the use of bilateral AFOs compared to walking in athletic shoes at 1.4m/s. [1] It has also been documented in healthy individuals that walking with a unilateral AFO results in reduced power generation at the ankle and increased extension moments at the hip on the braced side compared to shod walking at 1.2m/s. [9] While these findings provide valuable information on the behavior of the lower extremity joints when walking with reduced push-off, there still remain gaps in the literature. The differences in the trade-off at the ankle and hip between walking with unilateral reduced push-off and walking with bilateral reduced push-off have not been studied. Investigating the differences in the walking strategies between the various walking conditions is beneficial because it provides information on the likelihood of developing secondary conditions.

This study compared ankle and hip joint mechanics in 3 walking conditions (shod, unilateral brace and bilateral brace) using SAFOs. Sagittal plane ankle and hip angles, moments, and power were compared between three conditions separately for each side (braced side vs unbraced side) at 1.25m/s and 1.5m/s. We hypothesized that compared to shod walking, the reduced push-off from the use of the brace will decrease peak ankle angles, moments and powers and increase peak hip angles, moments and power. These changes were expected to increase with increasing speed.

2.2 Methods

2.2.1 Participants

Twelve healthy adults participated in this study between October 23,2020 and February 16, 2021. This sample size was selected using an *a priori* power analysis conducted in an opensource power analysis software (G*Power, Heinrich Heine Universität, Düsseldorf, Germany) assuming a large effect size of f=0.4, with an alpha level of 0.05 and a power of 0.8, 1 group and 3 measurements (using a repeated measures ANOVA within factors). The power analysis indicated that 12 participants were needed. Similar studies have also used similar sample sizes.[1, 9] Participants were included if they: a) were between the ages of 18 and 50 years; b) had a Body Mass Index (BMI) of less than 30 (this BMI criterion was used to prevent participants from deforming the brace as they walked and to reduce artifacts from soft tissues during data collections) and c) were comfortable with walking on a treadmill while wearing a SAFO. Participants were excluded if they self-reported: a) neuromuscular and/or musculoskeletal impairment(s); b) a history of cardiac or respiratory problems; c) a previous ankle fracture or surgery; d) lower extremity pain in the preceding three months; or e) required an assistive device to ambulate. (Table 1)

2.2.2 Instrumentation

Three-dimensional joint kinematics and vertical ground reaction forces were collected using a ten-camera motion capture system (Nexus 2.5, Vicon Motion Systems Ltd, Centennial, CO) sampling at 100Hz, and a split-belt treadmill with embedded force plates (Bertec Corporation, Columbus, OH) with a sampling frequency of 1000Hz.

2.2.3 Experimental Protocol

All research activities were approved by the Boston University Institutional Review Board. Data were collected in the Human Adaptation Laboratory, Sargent College, Boston University. Prior to the commencement of study activities, all participants were informed of the procedures and risks and signed an informed consent document. Each data collection visit lasted approximately two hours. Participants completed an intake form with questions on previous brace use and lower extremity injuries. Participant's shank measurements were recorded to select an appropriately sized AFO. Prefabricated 3/16-inch polypropylene AFOs with standard (sulcus length) foot plates (Optec USA) were used in this study.

The side on which the AFO was worn in the unilateral brace condition was randomized. Study participants wore spandex shorts, a form fitting shirt and their own pair of athletic shoes. Participants' weight and height were recorded, and their BMI calculated. Reflective spherical markers were placed bilaterally on the lower extremity, pelvis and trunk. Specifically, markers were placed over the acromion processes, spinous process of the seventh vertebrae, superior aspects of the iliac crests, anterior superior iliac spines (ASISs), posterior superior iliac spines (PSISs), sacrum, greater trochanters, lateral and medial femoral epicondyles, lateral and medial malleoli, calcaneus, and first and fifth metatarsal heads. Four plastic shells with four markers attached were secured to the thigh and shank segments with neoprene wraps.

2.2.4 Walking Conditions

Data were collected in three conditions and two speeds. In the shod condition, participants wore their own pair of athletic shoes. In the unilateral brace condition participants wore their athletic shoes and a prefabricated SAFO on a randomly selected limb and in the bilateral brace condition, participants wore prefabricated SAFOs and athletic shoes on both limbs. For all of these conditions, participants were asked to walk at a moderate speed of 1.25 m/s and a fast speed of 1.5 m/s. These speeds were chosen because 1.25 m/s represents an average walking speed and 1.5 m/s represents an increased walking speed. [21] Participants walked for at least two minutes in the three conditions at each speed.

2.2.5 Data Processing and Analysis

Motion data were processed in Vicon Nexus 2.5 (Vicon Motion Systems Ltd, Centennial, CO) and imported into Visual3D where joint angles were calculated using a hybrid model with a Cardan X-Y-Z rotation sequence. [22] Imported marker data were filtered using a low pass, fourth-order Butterworth filter with a cutoff frequency of 6 Hz. [23] Ground reaction force data were also low-pass filtered using a fourth-order Butterworth filter with a cutoff frequency of 10 Hz. [23] Internal joint moments and power were calculated using inverse dynamics based on kinematic marker positions and ground reaction force data. Push-off was quantified by ankle power generation in the sagittal plane: a variable which provides information on the energy generated per unit time. Sagittal plane joint angles, moments and power were calculated at the ankle and hip joints. Sagittal plane component of power was used. Kinetic data was exported to MATLAB (MathWorks, Natick, MA) for further analysis and transferred into Excel (Microsoft, Redmond, WA) for visual display.

2.2.6 Statistical Analysis

Data from the twelve participants were analyzed in SPSS (IBM, Chicago, IL) using generalized linear models (GLM) with general estimating equations (GEE). Our model included condition, speed, and the interaction between condition and speed. The independent variables in this study were the braced conditions and speed. The dependent variables were peak sagittal plane ankle and hip angles, moments and powers. We analyzed first and second peak hip power generation (H1 and H3) and power absorption during stance. Comparisons between conditions were done separately for each side and referred to as braced (ipsilateral to the brace) and unbraced (contralateral to the brace) sides for the unilateral brace condition. Sides in the shod and bilateral brace condition were named with the same reference as in the unilateral brace condition. Cohen's *d* effect sizes were calculated by dividing the mean by the standard deviations and interpreted as small (0.2), medium (0.5), and large (0.8). The significance criterion was set to 0.05.

2.3 Results

Results from GEE are presented in Table 2. Pairwise comparisons for variables with significant interaction effects between condition and speed are presented in Table 3.

Unbraced side, dorsiflexion angle: For peak dorsiflexion angle at the unbraced ankle joint, the GEE revealed no main effect for condition ($p=0.444$); however, there was a significant effect for speed ($p<0.001$). Peak dorsiflexion angle was 10% lower at the fast speed compared to the moderate speed (Mean difference=1.0º; Confidence Interval (CI)=0.7º,1.4º; *d*=1.9). There was no interaction effect between speed and condition

(p=0.117) on the unbraced side (Figure 1(a); Figure 2(a)).

Braced side, dorsiflexion angle: On the braced side, the GEE revealed no main effect for condition (p=0.300). There was a significant main effect for speed ($p<0.001$). Participants had 10% lower peak dorsiflexion angles at the fast speed compared to the moderate speed (Mean difference=1.1º; CI=0.8º,1.5º; *d*=2.2). There was no interaction effect between condition and speed (p=0.100) (Figure 1(b); Figure 2(b)).

Unbraced side, plantarflexion angle: For plantarflexion angle at the unbraced ankle joint, there were significant main effects for condition ($p<0.001$) and speed (p=0.004) but not for the interaction between condition and speed (p=0.387). Pairwise comparisons for condition on the unbraced side showed that compared to the shod condition, participants had 83% lower peak ankle plantarflexion angle in the bilateral brace condition ($p<0.001$; Mean difference=17.2º; CI=13.8º,20.7º; *d*=3.2). Compared to the unilateral brace condition, participants in the bilateral brace condition had 83% lower peak plantarflexion angle ($p<0.001$; Mean difference=16.6º; CI=13.9º,19.2º; *d*=4.1). On the unbraced side participants had 6% higher peak plantarflexion angles at the fast speed compared to the moderate speed (Mean difference=0.9º; CI=0.2º,1.5º; *d*=0.8) (Figure 1(a); Figure 2(c)).

Braced side, plantarflexion angle: On the braced side, the GEE revealed significant effects for condition ($p<0.001$) but not for speed (p=0.966) or the interaction between condition and speed (p=0.335). Pairwise comparisons for condition showed that compared to the shod condition, participants had 70% lower peak ankle plantarflexion angle in the unilateral brace condition ($p<0.001$; Mean difference=12.1º; CI=9.6º,14.5º; *d*=3.2) and 73% lower peak ankle plantarflexion angle in the bilateral brace condition ($p<0.001$; Mean difference=12.6º; CI=15.0º,10.3; *d*=3.5) (Figure 1(b); Figure 2(d)).

Unbraced side, dorsiflexion moments: For dorsiflexion moments, at the unbraced ankle joint, there were significant main effects for condition (p<0.001) and speed (p<0.001) but not for the interaction between condition and speed (p=0.401). Pairwise comparisons for condition on the unbraced side indicated that compared to the shod condition, participants had 87% higher peak ankle dorsiflexion moments in the bilateral brace condition (p<0.001; Mean difference=6.4Nm; CI=3.2Nm,9.6Nm; *d*=1.3). Also, compared to the unilateral brace condition, the bilateral brace condition had 84% higher peak ankle dorsiflexion moments (p<0.001; Mean difference=6.2Nm; CI=3.2Nm,9.3Nm; *d*=1.3). Peak ankle dorsiflexion moments were 13% higher at the fast speed compared to the moderate speed (p<0.001; Mean difference=1.2Nm; CI=0.6Nm,1.7Nm; *d*=1.4) (Figure 3(a); Figure 4(a)).

Braced side, dorsiflexion moments: On the braced side, the GEE revealed main effects for condition (p<0.001) and speed (p<0.001), as well as the for the interaction between speed and condition (p=0.036).

At the moderate speed, participants had 85% higher peak dorsiflexion moments in the unilateral brace (p<0.001; Mean difference=6.3Nm; CI=3.5Nm,9.2Nm; *d*=1.4) and 99% higher peak dorsiflexion moments in the bilateral brace (p<0.001; Mean difference=7.4Nm; CI=4.3Nm,10.5Nm; *d*=1.6) conditions compared to the shod condition. Participants had 8% higher peak dorsiflexion moments in the bilateral brace condition (p=0.042; Mean difference=1.1; CI=0.1Nm,2.3Nm; d=0.6) compared to the unilateral brace condition.

At the fast speed, participants had 70% higher peak dorsiflexion moments in the unilateral brace ($p<0.001$; Mean difference=6.0Nm; CI=3.5Nm,8.5Nm; d=1.6) and 88% higher peak dorsiflexion moments in the bilateral brace ($p<0.001$; Mean difference=7.6Nm; CI=4.5Nm,10.7Nm; d=1.6) conditions compared to the shod condition. Participants also had 11% higher peak dorsiflexion moments in the bilateral brace condition ($p=0.009$; Mean difference=1.6Nm; CI=0.2Nm,3.0Nm; d=0.8) compared to the unilateral brace condition (Figure 3(b); Figure 4(b)).

Unbraced side, plantarflexion moments: For plantarflexion moments at the unbraced ankle joint, there was no significant main effect for condition ($p=0.217$); however, the main effect for speed was significant ($p<0.001$). The interaction effect between condition and speed was not significant ($p=0.056$). Peak plantarflexion moments were 10% higher at the fast speed (Mean difference=9.7Nm; CI=7.7Nm,11.8Nm; d=3.1) compared to the moderate speed (Figure 3(a); Figure 4(c)).

Braced side, plantarflexion moments: On the braced side, the GEE revealed main effects for condition ($p<0.001$) and speed ($p<0.001$) but not for the interaction between condition and speed ($p=0.128$). Pairwise comparisons for condition showed that compared to the shod condition, participants had 3% lower peak plantarflexion moments in the unilateral brace condition ($p=0.012$; Mean difference=3.5Nm; CI=0.4Nm,6.6Nm; d=0.7) and 5% lower in the bilateral brace condition ($p<0.001$; Mean difference=5.2Nm; CI=2.1Nm,8.4Nm; d=1.1). Compared to the unilateral brace condition, participants had 2% lower peak ankle plantarflexion moments in the bilateral brace condition ($p=0.002$; Mean difference=1.8Nm; CI=0.5Nm,3.0Nm; d=0.9). Participants also had 10% higher peak plantarflexion moments at the fast speed compared to the moderate speed (Mean difference=10.0Nm; CI=8.3Nm,11.7Nm; *d*=3.9) (Figure 3(b); Figure 4(d)).

Unbraced side, ankle power generation: For power generation at the unbraced ankle joint, there were significant main effects for condition ($p<0.001$) and speed ($p<0.001$). The interaction between condition and speed was significant ($p<0.001$).

At the moderate speed, there were 3% lower peak ankle power generation in the unilateral brace ($p=0.022$; Mean difference=5.6Nm; CI=0.1Nm,11.0Nm; *d*=0.7) and 34% lower peak ankle power generation in the bilateral brace ($p<0.001$; Mean difference=71.7Nm; CI=52.3Nm,91.1Nm; *d*=2.4) conditions compared to the shod condition. Compared to the unilateral brace condition, participants had 32% lower peak ankle power generation in the bilateral brace condition ($p<0.001$; Mean difference=66.1Nm; CI=47.9Nm,84.3Nm; *d*=2.4).

At the fast speed, the difference in peak ankle power generation between the unilateral brace condition and the shod condition was not significant ($p=0.331$). Compared to the shod condition, participants had 34% lower peak ankle power generation in the bilateral brace condition ($p<0.001$; Mean difference=94.3Nm; CI=67.7Nm,120.9Nm; *d*=2.3). Participants also had 33% lower peak ankle power generation in the bilateral brace condition compared to the unilateral brace condition ($p<0.001$, Mean difference=91.1Nm; CI=67.2Nm,115.1Nm; *d*=2.5) (Figure 5(a); Figure 6(a)).

Braced side, ankle power generation: On the braced side, the GEE indicated significant main effects for both condition ($p<0.001$) and speed ($p<0.001$) but not for the interaction between condition and speed ($p=0.236$). Pairwise comparisons for condition showed that compared to the shod condition, participants had 31% lower peak power generation in the unilateral brace ($p<0.001$; Mean difference=70.7W; CI=49.6W,91.7W; d=2.2) and 33% lower peak power generation in the bilateral brace ($p<0.001$; Mean difference=76.5W; CI=56.4W,96.6W; d=2.5) conditions. Participants also had 28% higher peak ankle power generation at the fast speed compared to the moderate speed ($p<0.001$; Mean difference=44.0W; CI=33.3W,54.8W; d=2.7) (Figure 5(b); Figure 6(b)).

Unbraced side, ankle power absorption: For power absorption at the unbraced ankle joint, there were no main effects for condition ($p=0.782$) or speed ($p=0.945$). There was also no interaction effect between condition and speed ($p=0.613$) (Figure 5(a); Figure 6(c)).

Braced side, ankle power absorption: On the braced side, the GEE indicated no significant main effects for condition ($p=0.427$) or speed ($p<0.217$). There was also no interaction effect between condition and speed ($p=0.154$) (Figure 5(b); Figure 6(d)).

Unbraced side, hip flexion angle: For flexion angle at the hip joint on the unbraced side, there was no main effect for condition ($p=0.096$); however, there was an effect for speed ($p<0.001$). There was also no interaction effect between condition and speed ($p=0.306$). Peak hip flexion angle was 5% higher at the fast speed compared to the moderate speed (Mean difference=1.7^{o}; CI=1.1^{o},2.3^{o}; d=1.8) (Figure 7(a); Figure 8(a)).

Braced side, hip flexion angle: On the braced side, there were main effects for condition ($p<0.001$) and speed ($p<0.001$) but not for the interaction between condition and speed ($p=0.402$). Pairwise comparisons for condition showed that compared to the shod condition, participants had 7% higher peak hip flexion angles in the unilateral brace

($p<0.001$; Mean difference=2.4°; CI=1.6°,3.3°; *d*=1.9) and 4% higher peak hip flexion angles in the bilateral brace conditions ($p=0.020$; Mean difference= 1.4°; CI= 0.1°,2.8°; *d*=0.7). Participants had 3% lower peak hip flexion angles in the bilateral brace condition compared to the unilateral brace condition ($p=0.012$; Mean difference=1.0°; CI=0.1°,2.0°; *d*=0.7). Participants also had 5% higher peak hip flexion angle at the fast speed compared to the moderate speed (Mean Difference=1.8°; CI=1.2°,2.3°; *d*=2.2) (Figure 7(b); Figure 8(b)).

Unbraced side, hip extension angle: For hip extension angle on the unbraced side, the GEE indicated no main effect for condition ($p=0.636$); however, there was a main effect for speed ($p<0.001$). There was no interaction effect between speed and condition ($p=0.272$). The mean peak hip extension was 16% higher at the fast speed compared to the moderate speed (Mean difference=1.6°; CI=1.1°,2.1°; *d*=2.0) (Figure 7(a); Figure 8(c)).

Braced side hip, extension angle: On the braced side, the GEE revealed no main effect for condition ($p=0.906$) but a main effect for speed ($p<0.001$). There was no interaction effect between speed and condition ($p=0.642$). Compared to the moderate speed, participants had 18% higher mean peak hip extension at the fast speed (Mean difference= 1.6°; CI=1.2°,2.1°; *d*=2.4) (Figure 7(b); Figure 8(d)).

Unbraced side, hip flexion moments: For hip flexion moments on the unbraced side, the GEE indicated no significant main effect for condition ($p=0.864$) but the effect for speed was significant ($p<0.001$). There was no interaction effect between condition and speed ($p=0.174$). Participants had 22% higher peak hip flexion moments at the fast speed compared to the moderate speed (Mean difference=11.2Nm; CI=9.3Nm,13.0Nm; *d*=3.9)

(Figure 9(a); Figure 10(a)).

Braced side, hip flexion moments: On the braced side, there was no main effect for condition ($p=0.727$) but the main effect for speed was significant ($p<0.001$). There was no interaction effect between condition and speed ($p=0.584$). Compared to the moderate speed, participants had 21% higher peak hip flexion moments at the fast speed (Mean difference=10.5Nm; CI=8.9Nm,12.1Nm; d=4.3) (Figure 9(b); Figure 10(b)).

Unbraced side, hip extension moments: For hip extension moments on the unbraced side, there were main effects for condition ($p<0.001$) and speed ($p<0.001$) but no interaction effect between speed and condition ($p=0.365$). Pairwise comparison for condition showed that compared to the shod condition, participants had 8% higher peak hip extension in the unilateral brace condition ($p<0.001$; Mean difference=4.3Nm; CI=2.7Nm,5.8Nm; d=1.8). Compared to the unilateral brace condition, participants had 6% lower mean peak hip extension moments in the bilateral brace condition ($p=0.048$; Mean difference=3.1Nm; CI= 0.4Nm,6.5Nm; d=0.6). For the main effect for speed, compared to the moderate speed, participants had 28% higher peak hip extension at the fast speed (Mean difference=12.9Nm; CI=11.8Nm,14.0Nm; d=7.4) (Figure 9(a); Figure 10(c)).

Braced side, hip extension moments: On the braced side, the GEE indicated no effect for condition ($p=0.242$) but an effect for speed ($p<0.001$) and the interaction between speed and condition ($p=0.009$).

At the moderate speed, there were no significant differences in hip extension moments between the shod condition and the unilateral brace condition ($p=0.371$) or the bilateral brace condition ($p=0.695$). There was also no significant difference in hip extension moments between the unilateral brace condition and the shod condition (p=0.100).

At the fast speed, there were no significant differences in hip extension moments between the shod condition and the unilateral brace condition (p=0.647) or the bilateral brace condition (p=0.122). There was also no significant difference in hip extension moments between the unilateral brace condition and the shod condition (p=0.256) (Figure 9(b); Figure 10(d)).

Unbraced side, first peak hip power generation: For the first peak hip power generation on the unbraced side, there were significant effects for condition (p<0.001), speed (p<0.001) and the interaction between condition and speed (p=0.002).

At the moderate speed, there were 20% higher peak hip power generation in the unilateral brace (p<0.001; Mean difference=6.2W; CI=3.5W,9.0W; *d*=1.5) and 24% higher peak hip power generation in the bilateral brace (p<0.001; Mean difference=7.6W; CI=4.2W,11.1W; *d*=1.4) conditions compared to the shod condition. There was no significant difference between the unilateral and bilateral brace conditions (p=0.431).

At the fast speed, there were 20% higher peak hip power generation in the unilateral brace (p=0.010; Mean difference=8.7W; CI=1.1W,16.3W; *d*=0.7) and 32% higher peak hip power generation in the bilateral brace (p<0.001; Mean difference=13.9W; CI=6.1W,21.6W; *d*=1.2) conditions compared to the shod condition. Compared to the unilateral brace condition, participants had 10% higher peak hip power generation in the bilateral brace condition (p=0.001; Mean difference=5.2W; CI=1.8W,8.6W; *d*=1.0) (Figure 11(a); Figure 12(a)).

Braced side, first peak hip power generation: On the braced side, the GEE revealed significant main effects for condition ($p<0.001$), speed ($p<0.001$) and the interaction between condition and speed ($p=0.044$). At the moderate speed, compared to the shod condition, there were 17% and 21% higher peak hip power generation in the unilateral brace ($p<0.001$; Mean difference=6.3W; CI=2.7W,9.9W; d=1.1) and bilateral brace ($p<0.001$; Mean difference=7.6W; CI=3.7W,11.5W; d=1.3) conditions respectively. There were no significant differences between the unilateral brace and bilateral brace conditions ($p=0.465$).

At the fast speed, there were 22% higher peak hip power generation in the unilateral brace condition ($p<0.001$; Mean difference=11W; CI=5.2W,16.7W; d=1.2) and 24% higher peak hip power generation in the bilateral brace condition ($p<0.001$; Mean difference=12.1W; CI=7.2W,17.0W; d=1.6) compared to the shod condition. There were no significant differences between the unilateral brace condition and the bilateral brace condition ($p=0.659$) (Figure 11(b); Figure 12(b)).

Unbraced side, hip power absorption: For hip power absorption on the unbraced side, there was no effect for condition ($p=0.348$); however, there was an effect for speed ($p<0.001$). The was no interaction effect between speed and condition ($p=0.990$). Participants had 41% higher mean peak power absorption at the fast speed compared to the moderate speed (Mean difference=16.2W; CI=14.2W,18.3W; d=5.1) (Figure 11(a); Figure 12(c)).

Braced side, hip power absorption: On the braced side, there was no effect for condition ($p=0.378$) but there was an effect for speed ($p<0.001$). There was no interaction effect between condition and speed ($p=0.523$). Compared to the moderate speed, participants had 38% higher mean peak hip power absorption at the fast speed (Mean difference=14.7W; CI=12.6W,16.8W; d=4.5) (Figure 11(b); Figure 12(d)).

2.4 Discussion

Findings from this study support our hypotheses that the reduced push-off from solid ankle foot orthoses would result in decreased plantarflexion angles, plantarflexion moments and power generation at the ankle and increased hip angles, and peak power at the hip in both unilateral and bilateral brace conditions compared to the shod condition in the sagittal plane. These findings are consistent with the limb mechanics in the bipedal model [2] and are also consistent with findings from previous studies. [1, 9]

Contrary to our expectation however, dorsiflexion moments were increased with the use of SAFOs in both of the braced conditions. This finding can be attributed to the mechanical effects of the SAFO. The increased dorsiflexion moments observed when walking with decreased push-off is however consistent with findings from Lewis et al. [3]

On the unbraced side in the unilateral brace condition, there was decreased power generation at the ankle at the moderate speed. Extension moments, and first and second peak power generation at the hip were increased compared to the same side in the shod condition. The finding at the unbraced ankle can be attributed to the discrepancy in push-off between the two sides. Participants may have reduced push-off on the unbraced side to compensate for the behavior of the braced side. The increased peak hip extension moments observed at the moderate speed is also in agreement with the bipedal model of gait presented by Kuo et al. [2]

When comparing the unilateral brace condition to the bilateral brace condition, we found decreased plantarflexion moments and increased dorsiflexion moments on the braced side in the bilateral brace condition. Hip angles were also higher on the braced side in the bilateral brace condition. The observed differences in the trade-off between ankle and hip mechanics may be specific adaptations used when walking in each condition.

The analysis also revealed significant interaction effects between condition and speed for certain variables. Generally, the effect of the brace was larger at the fast speed compared to the moderate speed. This was observed for ankle dorsiflexion moments and first peak hip power generation on the braced side in both the unilateral and bilateral brace conditions. In the comparison between the two braced conditions, a larger effect of the brace was also observed at the fast speed for ankle power generation on the unbraced side and dorsiflexion moments on the braced side. This finding indicates that the effect of brace on gait increases with speed.

A similar study has been conducted on the mechanics and energetics of walking with bilateral reduced push-off on ankle and hip mechanics. [1] The use of bilateral ankle braces resulted in a decrease in peak ankle plantarflexion moments and power and an increase in extension moments and power at the hip joint. These findings from Huang et al. are mostly consistent with the findings from our study in the bilateral brace condition. Huang et al. also found increased extension moments with the use of the bilateral braces, but this was not observed in our study. A potential reason why this discrepancy was observed is how push-off was reduced. Compared to the thermoplastic SAFOs used in our study, Huang et al. used modified AFOs with steel cables attached across the ankle joint to restrict ankle plantarflexion. The use of the modified AFOs resulted in a much larger reduction in push-off than can be expected in real clinical scenarios and this may have resulted in the observed increase in extension moments at the hip joint. While we did not find an increase in extension moments in the bilateral condition, we observed an increase in hip extension moments in the unbraced limb in the unilateral brace condition.

A prior study conducted using unilateral solid ankle foot orthoses have also reported on kinetics at the hip and ankle joints. [9] Significant decreases in push-off and increases in extension moments and power were observed at the ankle and hip joints respectively in the braced limb. These findings are mostly consistent with findings from our study. We however did not find increased extension moments in the braced limb in the unilateral brace condition. The differences between this study and our study are the speed conditions and the braced conditions. Their study was conducted at slow (0.6 m/s) and moderate (1.2 m/s) speeds in a unilateral brace condition while this study was done at moderate (1.25 m/s) and fast (1.5 m/s) speeds in two braced conditions.

Lewis et al. have investigated the effects of bilateral increased and reduced push-off on lower limb mechanics. Contrary to our findings, Lewis et al. reported a significant difference between natural walking and walking with decreased push-off only for ankle dorsiflexion moments. This discrepancy can be attributed to how push-off was decreased in their study which was to instruct the participants to push-off less as they walked. [3] Since there were no external constraints to limit push-off or underlying pathology, participants walked in a similar way as they would when walking naturally, thus resulting in their findings.

Despite the energetic penalty associated with walking with reduced push-off, some studies have suggested that it may be advantageous to walk with decreased push off in certain populations. [24, 25] Individuals with diabetes mellitus benefit from walking with reduced push-off as this reduces the peak forefoot plantar pressures. [24] High forefoot plantar pressures are known to increase the likelihood of developing neuropathic foot ulcers which may result in amputations. Hence, it may be beneficial for individuals at high risk of developing foot ulcers to use modified (padded) AFO's together with their footwear. While this may be true, it is pertinent to ensure adequate hip muscle strength prior to the use of solid ankle foot orthoses as an alternative to reduce push-off especially when using SAFOs bilaterally.

The limitations of this study were the use of healthy participants and the use of prefabricated solid ankle foot orthoses. The use of participants without any neuromuscular or musculoskeletal conditions limits the application of the findings to only healthy individuals. Our findings are dependent on how push-off was reduced (i.e, using solid ankle foot orthoses). Our observations may have been different if a different type of brace was used to reduce push-off.

In conclusion, walking with SAFOs reduce push-off and changes hip mechanics. However, the compensations at the hip is different when walking with a unilateral brace compared to walking with bilateral SAFOs. Compared to walking with bilateral SAFOs, individuals increase extension moments at the hip in the unbraced limb when walking with a unilateral SAFO. The observed increased extension moments in the unbraced limb in the unilateral brace condition may increase an individual's risk of developing hip conditions such as hip pain, labral tears or hip instability.

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CHAPTER THREE

STUDY 2: EFFECT OF THE REDUCED PUSH-OFF FROM SOLID ANKLE FOOT ORTHOSES ON FRONTAL PLANE LOWER LIMB KINETICS.

3.1 Introduction

Walking with unilateral reduced push-off is a common problem observed when using a unilateral ankle foot orthosis (AFO). Individuals with lower extremity impairments such as multiple sclerosis or stroke are often prescribed unilateral AFOs to improve walking. [10, 26] While AFOs have been found to improve some parameters of pathological gait, [12, 27] the restriction of the ankle joint introduced by the brace also adds to the problems with push-off from the conditions. Reduced push-off during gait is inefficient and has been associated with altered lower extremity mechanics and increased energy expenditure. [1, 17] Furthermore, individuals who walk with unilateral reduced push-off abnormally load their contralateral limb; a compensation which has been related to the development of secondary conditions such as osteoarthritis (OA) and pain in their unaffected limb. [5, 6, 28, 29]

Increased ground reaction forces leading to atypical joint loading have been linked to the development of OA in the unaffected limb. [29, 30] At the knee joint, increased internal knee abduction moments result in increased stresses on the medial compartment of the knee, which is understood to be a principal factor in the development, severity and progression of OA. [5, 28, 31] At the hip joint, internal abduction moments have been positively correlated with femoral neck bone mineral density [28, 30, 32] in individuals with hip OA. Previous studies in unilateral transtibial amputees have shown that these individuals have a higher risk [5] and prevalence [6] of developing OA in their non-amputated limb. [29, 33] While the underlying biomechanical mechanisms for these relationships with OA is unclear, reduced push-off is known to play a role (Morgenroth, et al., 2011; Silverman & Neptune, 2014). [29, 34] These previous studies have only reported on the reduced push-off associated with the use of unilateral transtibial prostheses. However, individuals that ambulate with unilateral AFOs also experience reduced push-off in the braced limb, exhibit similar gait mechanics as unilateral transtibial prosthesis users, [9] and thus may be at an increased risk of developing OA in their unbraced limb.

The aim of this study was to evaluate the effects of the reduced push-off associated with the use of unilateral solid AFOs (SAFOs) on the kinetics of the unbraced lower limb in healthy individuals. We hypothesized that, compared to shod walking, the use of a unilateral SAFO would result in an increase in peak internal abduction moments at the knee and hip joints and an increase in peak vertical ground reaction forces in the unbraced limb.

3.2 Methods

3.2.1 Participants

Twelve healthy adults participated in this study between October 23, 2020 and February 16, 2021. This sample size was selected using an *a priori* power analysis conducted in an open source power analysis software (G*Power Heinrich Heine Universität, Düsseldorf, Germany) assuming a medium effect size d=0.50, an alpha level of 0.05 and a power of 0.80 with a one tailed paired sample T-test with matched pairs. The power analysis indicated that 27 participants were needed. Twelve participants were used in this study due to restrictions on research activities during the COVID-19 pandemic. Participants were included if they: a) were between the ages of 18 and 50 years; b) had a

Body Mass Index (BMI) of less than 30 (this BMI criteria was used to prevent participants from deforming the brace as they walked and to reduce artifact from soft tissues) and c) were comfortable with walking on a treadmill while wearing a SAFO. Participants were excluded if they self-reported: a) neuromuscular and/or musculoskeletal impairment(s); b) a history of cardiac or respiratory problem(s); c) a previous ankle fracture or surgery; d) lower extremity pain in the preceding three months; or e) required an assistive device to ambulate. (Table 1)

3.2.2 Instrumentation

Three-dimensional joint kinematics and ground reaction forces were collected using a ten-camera motion capture system (Nexus 2.5, Vicon Motion Systems Ltd, Centennial, CO) sampling at 100 Hz, and a split-belt treadmill embedded with force plates (Bertec Corporation, Columbus, OH) sampling at 1000 Hz.

3.2.3 Experimental Protocol

All research activities were approved by the Boston University Institutional Review Board. Data were collected in the Human Adaptation Laboratory, Sargent College, Boston University. Each data collection visit lasted approximately two hours. Prior to the commencement of study activities, all participants were informed of the procedures and risks and signed an informed consent document. Participants completed an intake form with questions on previous brace use and lower extremity injuries. Participants' shank measurements were recorded to select an appropriately sized AFO. Prefabricated 3/16-inch polypropylene AFOs with sulcus length standard foot plates (Optec USA) were used in this study.

Since participants were healthy individuals, the side on which to wear the AFO in the unilateral brace condition was randomized. Study participants wore spandex shorts, a form fitting shirt and their own pair of athletic shoes. Participants' weight and height were recorded, and their BMI calculated. Reflective spherical markers were placed bilaterally on the lower extremity, pelvis and trunk. Specifically, markers were placed over the acromion processes, spinous process of the seventh vertebrae, superior aspects of the iliac crests, anterior superior iliac spines (ASISs), posterior superior iliac spines (PSISs), greater trochanters, lateral and medial femoral epicondyles, lateral and medial malleoli, calcaneus, and first and fifth metatarsal heads. [35] Four plastic shells with four markers attached collinearly were secured to the thigh and shank segments with neoprene wraps.

3.2.4 Interventions

Data were collected in two conditions and two speeds. In the shod condition, participants wore their own pair of athletic shoes. In the unilateral brace condition participants wore their athletic shoes and a prefabricated SAFO on a randomly selected limb. For both of these conditions, participants were asked to walk at a moderate speed of 1.25m/s and a fast speed of 1.5m/s. These speeds were chosen because 1.25m/s represents an average walking speed and 1.5m/s represents an increased walking speed. [21] Participants walked for at least two minutes in the two conditions at each speed.

3.2.5 Data Processing and Analysis

Motion data were labeled, interpolated and gap filled in Vicon Nexus 2.5 (Vicon

Motion Systems Ltd, Centennial, CO) and imported into Visual3D (C-motion Inc., Germantown, MD) where they were filtered using a low pass, fourth-order Butterworth filter with a cutoff frequency of 6Hz. [23] Joint angles were calculated in Visual3D using a hybrid model with a Cardan X-Y-Z rotation sequence. [22] Ground reaction force data were also low-pass filtered using a fourth-order Butterworth filter with a cutoff frequency of 10 Hz. [23] Internal joint moments and power were calculated using inverse dynamics based on kinematic marker positions and ground reaction force data. Push-off was quantified by ankle power generation in the sagittal plane. This variable was chosen because it provides information on the energy generated per unit time. Frontal plane joint moments were calculated at the hip and knee joints as well as vertical ground reaction forces. Kinetic data were exported to MATLAB (MathWorks, Natick, MA) for further analysis and transferred into Excel (Microsoft, Redmond, WA) for visual display.

3.2.6 Statistical Analysis

Data from the twelve participants were analyzed in SPSS (IBM, Chicago, IL) using one tailed paired sample t-tests. The independent variables in this study were walking condition and speed. The dependent variables were frontal plane knee and hip abduction moments and vertical ground reaction forces. Separate analyses were run for each speed. We analyzed the first peak for each dependent variable because it represented the highest peak and has been the peak that has been linked with OA. [36] The Cohen's *d* effect sizes were calculated using the mean and standard deviations of the dependent variables and interpreted as small (0.2), medium (0.5), and large (0.8). [37] The significance criterion was set to 0.05.

3.3 Results

Ankle Power Generation: Compared to shod walking, the use of the SAFO resulted in 29.79% and 31.6% reduction in peak ankle power at the moderate ($p<0.001$; Mean difference=64.99W; Confidence Interval (CI)=40.8,89.2W; d=2.0) and the fast ($p<0.001$; Mean difference=76.4W; CI=45.8,107.0W; d=1.9) speeds respectively in the braced limb (Table 4).

Knee Moments: At the moderate speed, the first peak internal knee abduction moment was 3% higher ($p=0.041$) for the unbraced limb in the braced condition compared to the corresponding limb in the shod condition (Mean difference=0.9Nm; CI=-0.1,1.8; d=0.7). There were no significant differences in the first peak internal knee abduction moments ($p=0.600$) at the fast speed (Figure 13).

Hip Moments: The first peak internal hip abduction moment was 4% higher ($p=0.002$) for the unbraced limb in the braced condition compared to the corresponding limb in the shod condition at the moderate speed. (Mean difference=2.6Nm; CI=4.3,4.1; d=1.2). At the fast speed, there were no significant differences in hip abduction moments ($p=0.070$) (Figure 14).

Vertical ground reaction force: At the moderate speed, the peak vertical ground reaction force was 3% higher ($p=0.002$) in the unbraced limb in the unilateral brace condition compared to the corresponding limb in the shod condition (Mean difference=18.6N; CI=7.0,30.1; d=1.2). Peak vertical ground reaction force was 3% higher ($p=0.002$) for the unbraced limb in the braced condition compared to the corresponding limb in the shod condition at the fast speed (Mean difference of 21.0N; CI=8.0-34.0; d=1.2)

(Figure 15).

(Table 5)

3.4 Discussion

Findings support our hypothesis that the reduced push-off associated with the use of a unilateral SAFO results in increased joint kinetics in the unbraced limb. We found increased internal knee and hip abduction moments and vertical ground reaction forces in the unbraced limb when walking with the unilateral SAFO compared to walking with athletic shoes. This finding is consistent with previous studies on the effect of unilateral reduced push-off on joint kinetics. [28, 34] These previous studies, however, have only reported on the reduced push-off associated with the use of unilateral transtibial prostheses.

Knee abduction moments have been investigated in similar studies in unilateral transtibial amputees. For instance, Royer et al. have reported 10% and 17% greater peak internal knee abduction moments in the intact limb of transtibial amputees when compared with controls. [28, 30] Some of these studies have compared the knee abduction moments between the amputated limb and the intact limb and have found significant differences between the two sides with the prosthetic side having a lower peak than the sound side. [28] These differences between sides may not mean the intact limb is loaded more if it is not compared to controls.

Clinically significant values for knee abduction moments have been reported to be about 5-7% . [34, 38] While the effects we found were slightly below those thresholds for clinically meaningful changes, this needs to be interpreted considering the fact that healthy participants were used in our study. Since the participants in our study had intact ankle joint structures and functions and unaltered muscle strength, they were likely to deform the SAFOs when pushing-off more than individuals with musculoskeletal conditions can. Accordingly, individuals with musculoskeletal conditions can be expected to exhibit more severely reduced push-off on the braced side and this may result in a greater increase in the knee and hip abduction moments.

Our study also found large and medium effects of 1.19 and 0.58 representing a 4% and 3% increase in frontal plane hip moments at the moderate and fast speeds respectively in the unbraced limb when walking with a unilateral SAFO compared to the same side during shod walking. This is in comparison to the 23% difference reported by Royer et al. in unilateral transtibial amputees. [30] Another study has reported 11% higher peak hip abduction moments at a slow speed and 16% higher peak hip abduction moments at a moderate speed in the unbraced limb when using a unilateral SAFO compared to shod walking. [9] While our findings are similar, the magnitude of the differences seen were not as high. A possible reason for the discrepancy between the two studies was the SAFO used in their study. The full foot plate and ankle straps on the SAFO used in Vistamehr et al. may have resulted in more restriction of the ankle joint and a greater reduction in push-off generation compared to the sulcus length footplate used in our study. The greater reduction in push-off may account for the larger difference in hip abduction moments between the unilateral brace condition and shod condition. The study by Vistamehr et al. also reported the average peak hip abduction moments values of both limbs in the shod condition. By doing this, they assumed symmetric behavior of the two limbs in the shod condition. Our study compared the unbraced limb in the unilateral brace condition to the corresponding side in the shod condition.

Another factor that may contribute to increased joint loading is vertical ground reaction force. A study has reported higher vertical ground reaction forces in the intact limb of unilateral transtibial amputees when compared to healthy controls. [33] Findings from these studies are consistent with findings from our study when comparing the unbraced limb in the unilateral brace condition to shod walking but the magnitude of the difference observed in our study was smaller. The differences in the magnitudes observed in the two studies can be attributed to how push-off was reduced. The use of unilateral transtibial prostheses is expected to reduce push-off to a greater extent than walking with a unilateral SAFO.

Due to the large observed effects of SAFOs on push-off, SAFOs should be reserved for conditions that cannot be treated otherwise. Clinicians should first consider using other types of AFOs such us energy returning carbon fiber AFOs which can contribute to push-off and reduce the likelihood of developing secondary conditions such as OA. In the case that a unilateral SAFO is an individual's best treatment option, clinicians may need to treat the unbraced limb as well. Clinicians may also consider requesting to supplement the prescription for an SAFO with lateral shoe wedges for the intact limb since lateral wedges are known to offload the medial knee joint and reduce the risk of developing knee OA. [38] Also, gait retraining interventions to manage the bilateral discrepancy in push-off may be beneficial.

A limitation of this study is the use of healthy, asymptomatic individuals. This helped control for many confounding variables typically associated with a symptomatic population. The use of prefabricated SAFOs may not represent what is usually prescribed in clinics; however, we do not anticipate these differences to be large since the prefabricated braces used in our study had a similar thickness as the custom molded orthoses dispensed in clinics. The sample size used in this study as well as the limited adaptation time are also potential limitations.

In conclusion, the reduced push-off associated with the use of unilateral SAFOs result in significant changes in frontal plane kinetics at the knee and hip of the unbraced limb. The increased internal knee and hip abduction moments and vertical ground reaction force in the unbraced limb may increase an individual's likelihood of developing OA, especially with long term use.

CHAPTER FOUR

4.1 Conclusion

Findings from the first study show that different compensations are used at the hip when walking with unilateral reduced push-off compared to walking with bilateral reduced push-off. Differences in reduced push-off conditions were compensated by increasing extension moments and power generation at the hip in the unilateral brace condition and increasing power generation only in the bilateral brace condition. The observed differences in the compensations at the hip in the unilateral brace condition varied between the braced and unbraced sides.

On the unbraced side, we found a decrease in push-off in the unilateral brace condition compared to the shod condition at the moderate speed as an adaptation to the reduced push-off found in the contralateral limb. The changes in the ankle mechanics were compensated for at the hip joint by increasing hip kinetics. Participants had higher peak hip extension and first and second peak power generation at the hip in the unilateral brace compared to the shod condition and higher first and second peak hip power generation in the bilateral brace condition compared to the shod condition

On the braced side, the decreased kinetics at the ankle joint were compensated for at the hip joint by increasing only hip power generation in both braced conditions. First and second peak hip power generation was higher in both the unilateral brace and bilateral brace condition compared to the shod condition. No differences were observed in hip kinetics in both braced conditions.

Based on these findings we can conclude that, there are different trade-offs between ankle and hip mechanics when walking with bilateral compared to walking with unilateral reduced push-off. When walking with unilateral reduced push-off, individuals increase the extension moments in the unbraced limb which may lead to hip conditions such as anterior hip pain in the unbraced limb.

In the second study, the use of the unilateral brace resulted in increased peak frontal plane moments at the knee and hip as well as vertical ground reaction forces compared to shod walking. Based on these findings we can conclude that the reduced push-off from the use of unilateral solid ankle foot orthoses increases an individual's risk of developing knee and hip osteoarthritis.

In conclusion, the findings from this dissertation show that walking with unilateral reduced push-off increases the likelihood of developing secondary conditions compared to walking with bilateral reduced push-off. While findings supported our hypotheses, there were still some limitations to these studies. The healthy participants and the small sample sizes were limitations to this study. Also, the prefabricated brace and the short adaptation time before data collections were also potential limitations to this study. Future work could examine these differences using a larger sample size. Increasing the sample size may reveal significant findings that may have not been observed with the small sample size used in these studies. Future work could also investigate the effects unilateral and bilateral reduced push-off on the energetics of walking as the trade-offs between ankle and hip kinetics are not the same.

APPENDIX

Tables

Participant demographics (N=12)	**Mean**	**SD**
Age (years)	25.1	3.6
Height (m)	1.69	0.07
Mass (kg)	66.0	8.6
BMI (kg/m^2)	23.1	

Table 1*: Participant demographics*

Main effects	Shod Condition (SC)		Unilateral brace (UB)		Bilateral brace (BB)		Cohen *d*'s Effect size			Statistical analysis		
							SC vs UB	SC vs BB	UB vs BB			
	Mean	SD	Mean	SD	Mean	SD				t	df	p
Unbraced side												
Peak Angle(º)												
Ankle dorsiflexion	9.42	2.7	9.92	3.4	10.92	5.1	0.29	0.36	0.29	1.6	2	0.444
Ankle plantarflexion	20.68	4.3	20.00	4.4	3.43	5.0	-0.30	**-3.22**	**-4.08**	222.2	2	< 0.001
Hip flexion	33.68	6.2	34.17	6.3	35.49	6.5	0.36	**0.62**	**0.59**	4.7	2	0.096
Hip extension	11.49	7.0	10.97	7.4	10.94	7.9	-0.27	-0.20	-0.01	0.9	2	0.636
Peak Joint Moment (Nm)												
Ankle dorsiflexion	7.32	2.5	7.46	2.9	13.70	6.1	0.14	**1.30**	**1.31**	20.7	2	<0.001
Ankle plantarflexion	106.99	17.1	106.46	16.7	103.70	19.2	-0.28	-0.43	-0.36	3.1	2	0.217
Hip flexion	56.68	10.5	55.90	10.7	55.77	12.0	-0.15	-0.15	-0.04	0.3	2	0.864
Hip extension	50.96	6.7	55.23	6.8	52.17	8.9	**1.84**	0.22	**-0.57**	41.1	2	<0.001

Peak Joint Power (W)												
Ankle power generation	247.16	49.7	242.75	47.0	164.12	36.0	-0.52	**-2.37**	**-2.48**	73.9	2	<0.001**
Ankle power absorption	56.96	20.1	58.65	18.3	57.37	21.5	0.20	0.04	-0.12	0.5	2	0.782
Hip power generation (H1)	37.29	12.7	44.74	14.1	48.03	14.7	**1.12**	**1.42**	**0.63**	24.2	2	<0.001**
Hip power generation (H3)	70.09	9.1	73.68	8.6	79.32	13.0	**0.58**	**0.80**	**0.65**	7.8	2	0.020
Hip power absorption	49.51	11.5	46.47	10.5	46.95	11.4	-0.42	-0.32	0.10	2.1	2	0.348
Braced Side												
Peak Angle(°)												
Ankle dorsiflexion	10.96	3.6	11.17	3.4	10.37	3.2	0.06	-0.26	-0.34	2.4	2	0.300
Ankle plantarflexion	17.36	4.5	5.29	2.1	4.71	1.8	**-3.19**	**-3.45**	-0.41	142.6	2	<0.001
Hip flexion	35.75	6.2	38.19	6.7	37.15	6.4	**1.89**	**0.67**	**-0.72**	56.4	2	<0.001
Hip extension	9.92	7.4	9.73	8.1	9.89	8.0	-0.10	-0.01	0.08	0.2	2	0.906

Peak Joint Moment (Nm)												
Ankle dorsiflexion	8.06	1.8	14.24	4.9	15.58	5.6	**1.52**	**1.59**	**0.69**	30.6	2	<0.001**
Ankle plantarflexion	104.56	17.5	101.08	17.4	99.32	16.5	**-0.73**	**-1.06**	**-0.90**	18.5	2	<0.001
Hip flexion	55.49	8.7	56.15	9.5	55.34	9.4	0.19	-0.03	-0.18	0.6	2	0.727
Hip extension	52.41	6.6	52.17	9.4	53.83	8.9	-0.04	0.30	0.42	2.8	2	0.242**
Peak Joint Power (W)												
Ankle power generation	231.05	44.5	160.36	43.1	154.51	32.6	**-2.17**	**-2.46**	-0.22	79.2	2	<0.001
Ankle power absorption	67.7	25.5	72.12	30.0	67.70	23.5	0.19	0.00	-0.36	1.7	2	0.427
Hip power generation (H1)	43.67	11.6	52.28	13.3	53.50	12.4	**1.38**	**1.66**	0.18	40.2	2	<0.001**
Hip power generation (H3)	78.41	8.8	85.69	8.6	84.50	11.4	**0.98**	**0.86**	-0.20	12.6	2	0.002
Hip power absorption	45.55	13.7	47.56	14.0	46.15	11.9	0.39	0.08	-0.21	1.94	2	0.378

Table 2: Main effect of condition for braced and unbraced sides (**indicates a significant interaction effect between speed and condition.)

Ankle dorsiflexion moments (Nm) (Braced side)		Mean difference	Percentage change (%)	SD	Cohen d's Effect size	df	p
UB*fast	SC*fast	6.02	70	3.82	1.58	1	<0.001
BB*fast	SC*fast	7.62	88	4.80	1.59	1	<0.001
BB*fast	UB*fast	1.60	11	2.12	0.75	1	0.009
UB*moderate	SC*moderate	6.33	85	4.42	1.43	1	<0.001
BB*moderate	SC*moderate	7.43	99	4.79	1.55	1	<0.001
BB*moderate	UB*moderate	1.09	8	1.86	0.59	1	0.042
Ankle power generation (W) (Unbraced side)							
UB*fast	SC*fast	-3.22	-1	11.49	-0.28	1	0.331
BB*fast	SC*fast	-94.34	-34	41.11	-2.29	1	<0.001
BB*fast	UB*fast	-91.12	-33	37.02	-2.46	1	<0.001
UB*moderate	SC*moderate	-5.59	-3	8.43	-0.66	1	0.022
BB*moderate	SC*moderate	-71.73	-34	29.99	-2.39	1	<0.001
BB*moderate	UB*moderate	-66.14	-32	28.15	-2.35	1	<0.001

Hip extension moments (Nm) (Braced side)							
UB*fast	SC*fast	0.85	1	6.45	0.13	1	0.647
BB*fast	SC*fast	2.30	4	5.15	0.45	1	0.122
BB*fast	UB*fast	1.45	2	4.42	0.33	1	0.256
UB*moderate	SC*moderate	-1.33	3	5.17	-0.26	1	0.371
BB*moderate	SC*moderate	0.54	1	4.72	0.11	1	0.695
BB*moderate	UB*moderate	1.87	4	3.93	0.48	1	0.100
First peak hip power generation (W) (Unbraced Side)							
UB*fast	SC*fast	8.67	20	11.7	0.74	1	0.010
BB*fast	SC*fast	13.85	32	12.0	1.15	1	<0.001
BB*fast	UB*fast	5.19	10	5.3	0.98	1	0.001
UB*moderate	SC*moderate	6.24	20	4.3	1.45	1	<0.001
BB*moderate	SC*moderate	7.63	24	5.3	1.44	1	<0.001

BB*moderate	UB*moderate	1.39	4	6.1	0.23	1	0.431
First peak hip power generation (W) (Braced side)							
UB*fast	SC*fast	10.96	22	8.9	1.23	1	<0.001
BB*fast	SC*fast	12.08	24	7.6	1.59	1	<0.001
BB*fast	UB*fast	1.12	2	8.8	0.13	1	0.659
						1	
UB*moderate	SC*moderate	6.27	17	5.5	1.14	1	<0.001
BB*moderate	SC*moderate	7.58	21	6.0	1.26	1	<0.001
BB*moderate	UB*moderate	1.31	3	6.2	0.21	1	0.465

Table 3*: Pairwise comparisons for significant interaction effects.*

	Shod condition	**SD**	**Unilateral brace condition**	**SD**	**Cohen's D effect size**	**Percentage change**	**Test statistic**	**P-value**
Ankle power generation (W)								
Moderate speed (1.25m/s)	205.65	36.8	140.66	41.3	2.01	-29.79%	6.97	**<0.001**
Fast speed (1.5m/s)	256.44	59.1	180.05	49.8	1.87	-31.6%	6.48	**<0.001**

Table 4*: Peak ankle power generation (braced side).*

	Shod condition	SD	Unilateral brace condition	SD	Cohen d's effect size	Percentage change (%)	Statistical analysis		
							t	df	p
Moderate speed (1.25m/s)									
Internal knee abduction moments (Nm)	28.46	6.82	29.33	7.23	0.67	3	2.31	11	**0.041**
Internal hip abduction moments (Nm)	60.46	10.13	63.08	10.09	1.19	4	4.12	11	**0.002**
Vertical ground reaction forces (N)	739.88	99.47	758.47	99.93	1.20	3	4.17	11	**0.002**
Fast speed (1.5m/s)									
Internal knee abduction moments (Nm)	31.62	8.07	31.88	8.47	0.16	1	0.54	11	0.599
Internal hip abduction moments (Nm)	66.37	10.83	68.58	11.03	0.58	3	2.01	11	0.070
Vertical ground reaction forces (N)	796.29	105.07	817.31	106.54	1.21	3	4.18	11	**0.002**

Table 5: Peak knee and hip internal abduction moments and ground reaction forces at fast and moderate speeds (unbraced side).

Figures

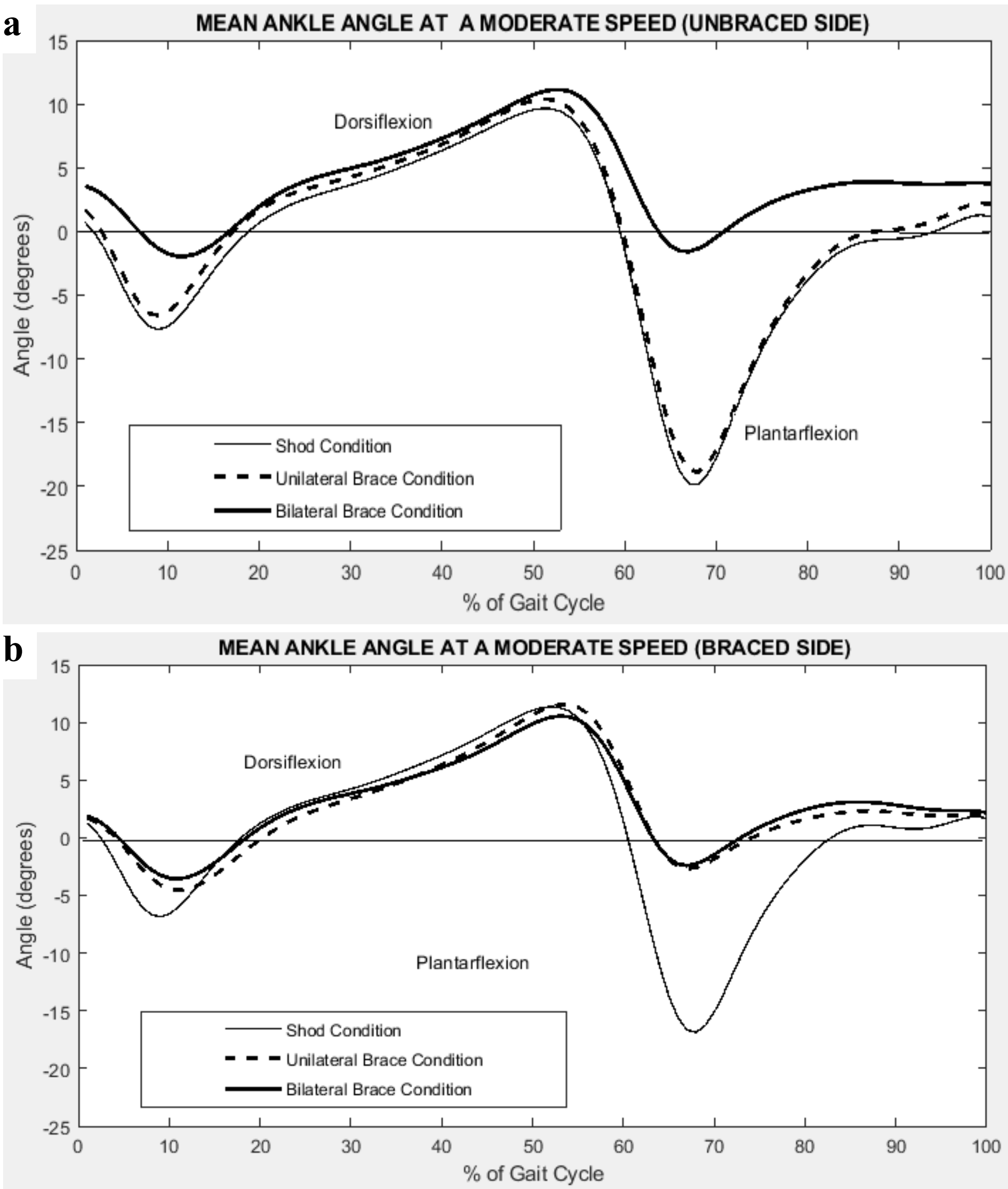


Figure 1*: Mean ankle angle at a moderate speed on the (a) unbraced and (b) braced sides.*

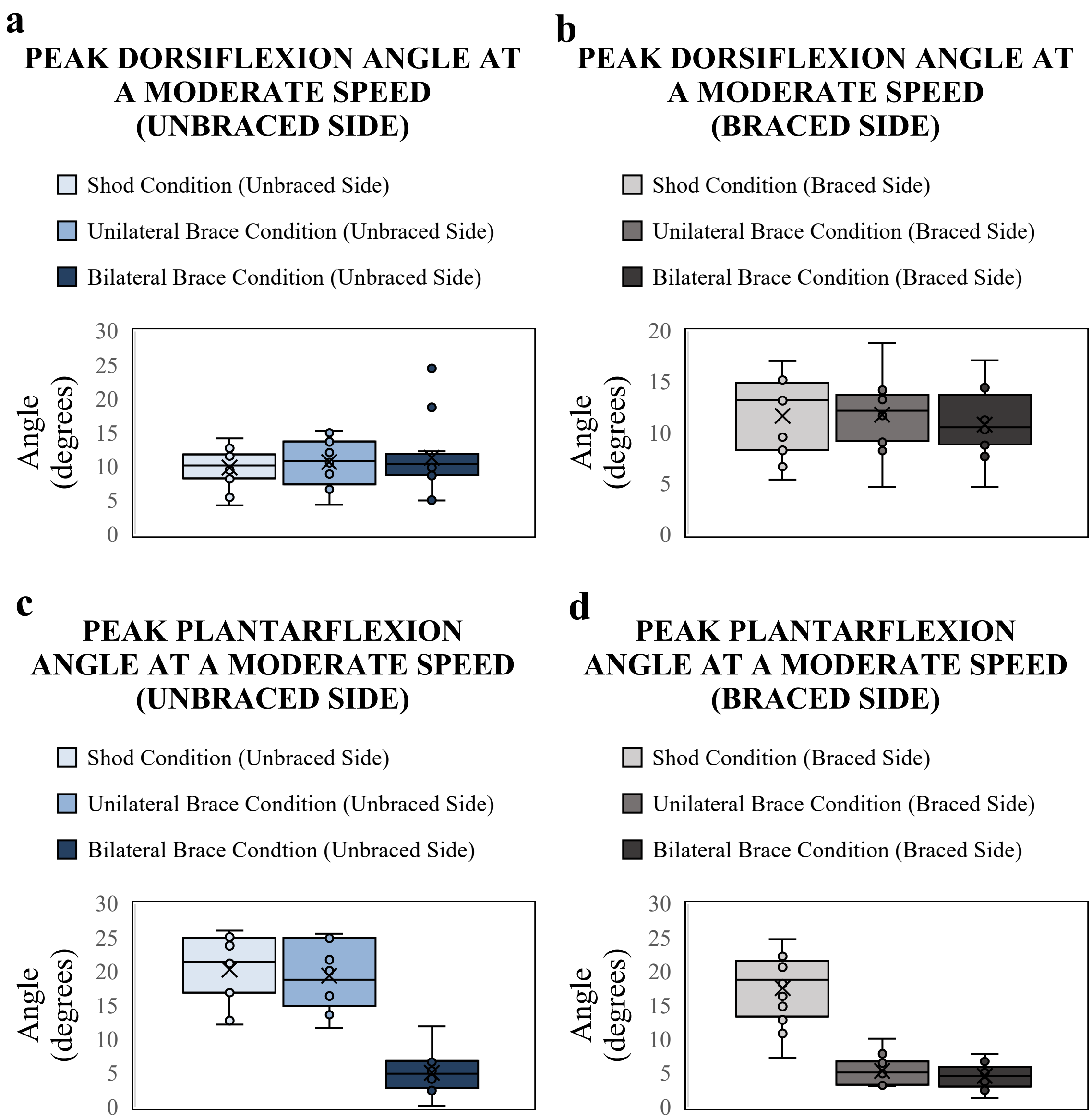


Figure 2*: Peak ankle dorsiflexion ((a) unbraced, (b) braced) and plantarflexion ((c) unbraced, (d) braced) angle at a moderate speed.*

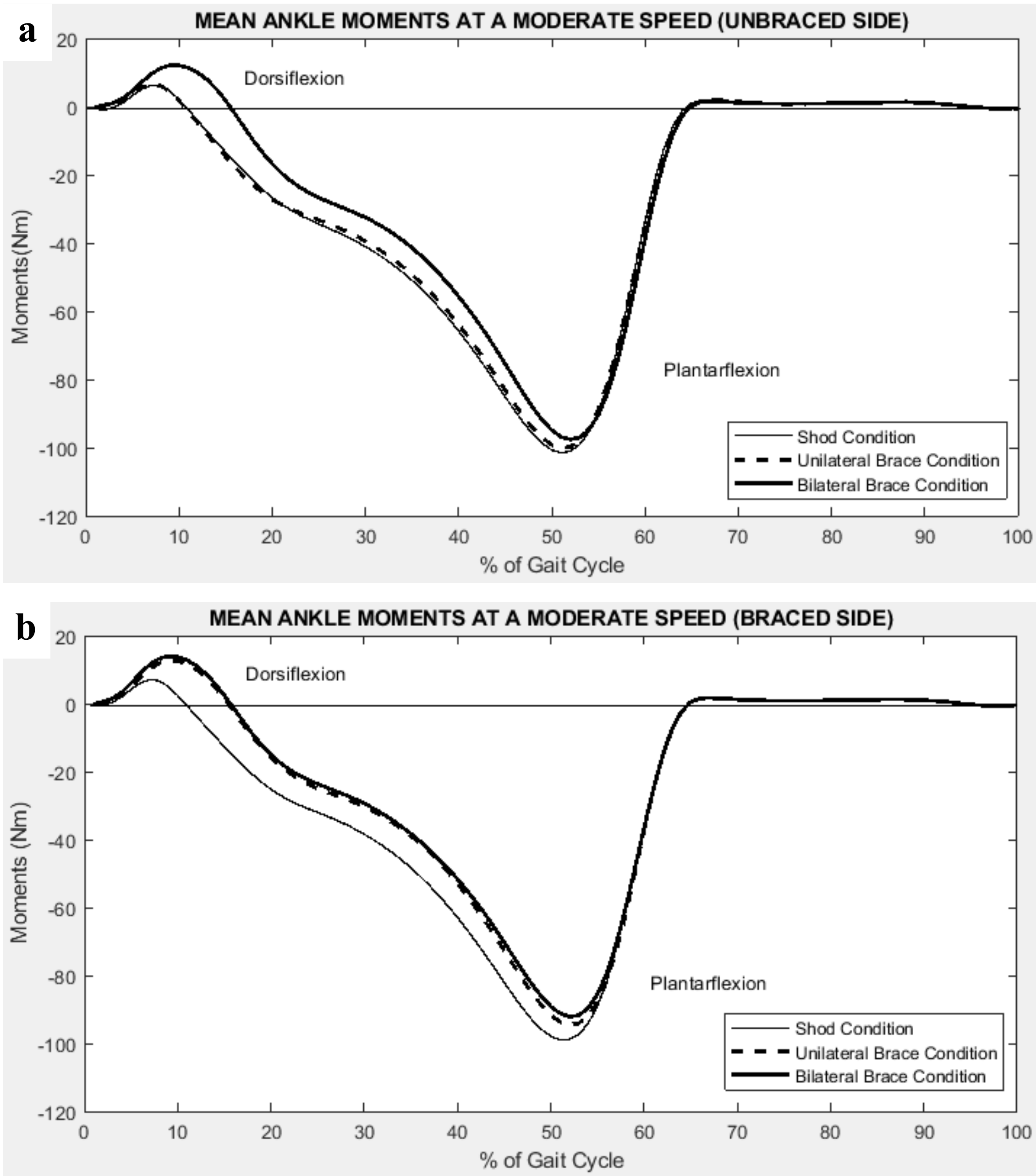


Figure 3*: Mean ankle moments at a moderate speed on the (a) unbraced and (b) braced sides.*

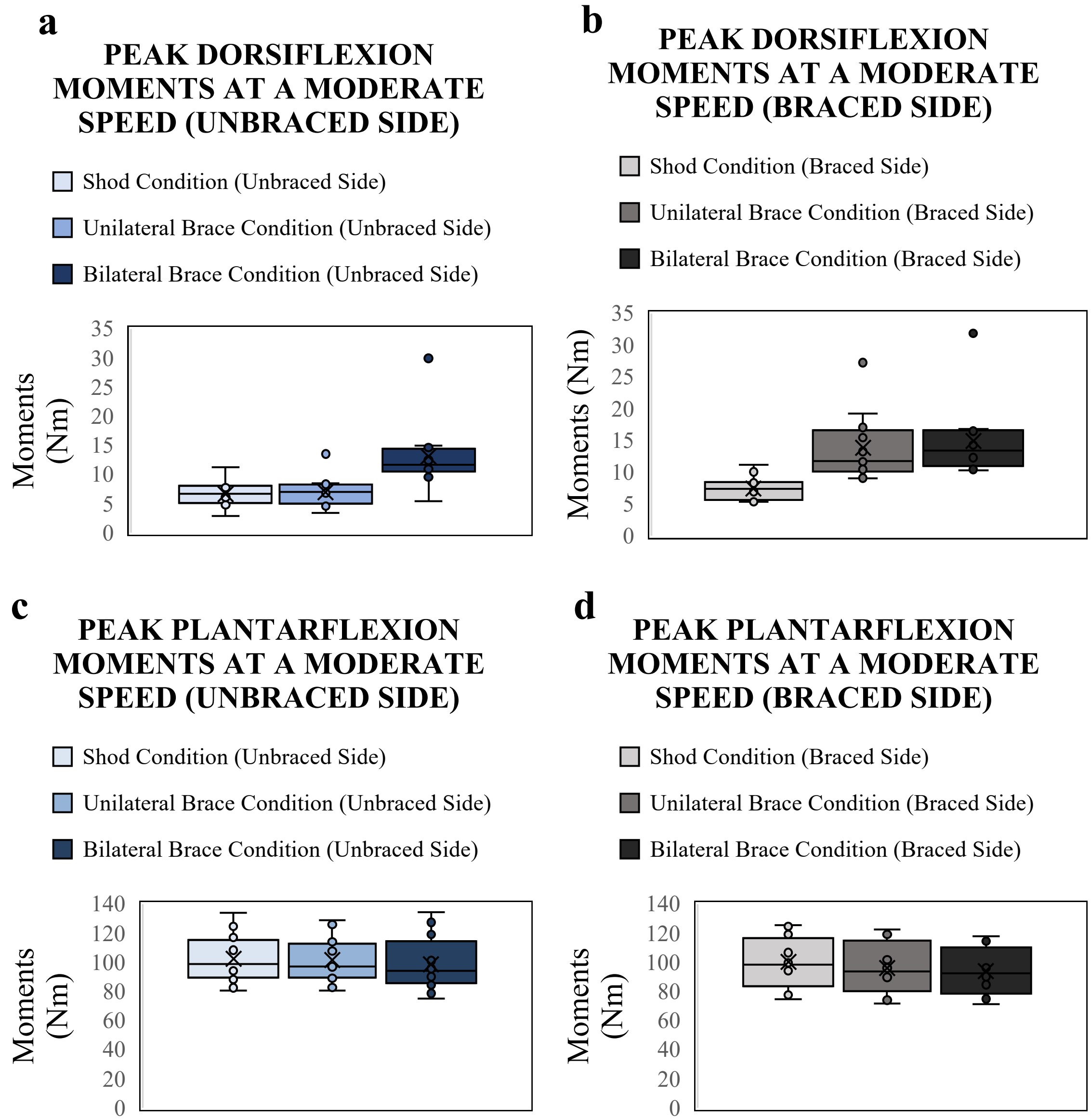


Figure 4*: Peak ankle dorsiflexion ((a) unbraced, (b) braced) and plantarflexion ((c) unbraced, (d) braced) moments at a moderate speed.*

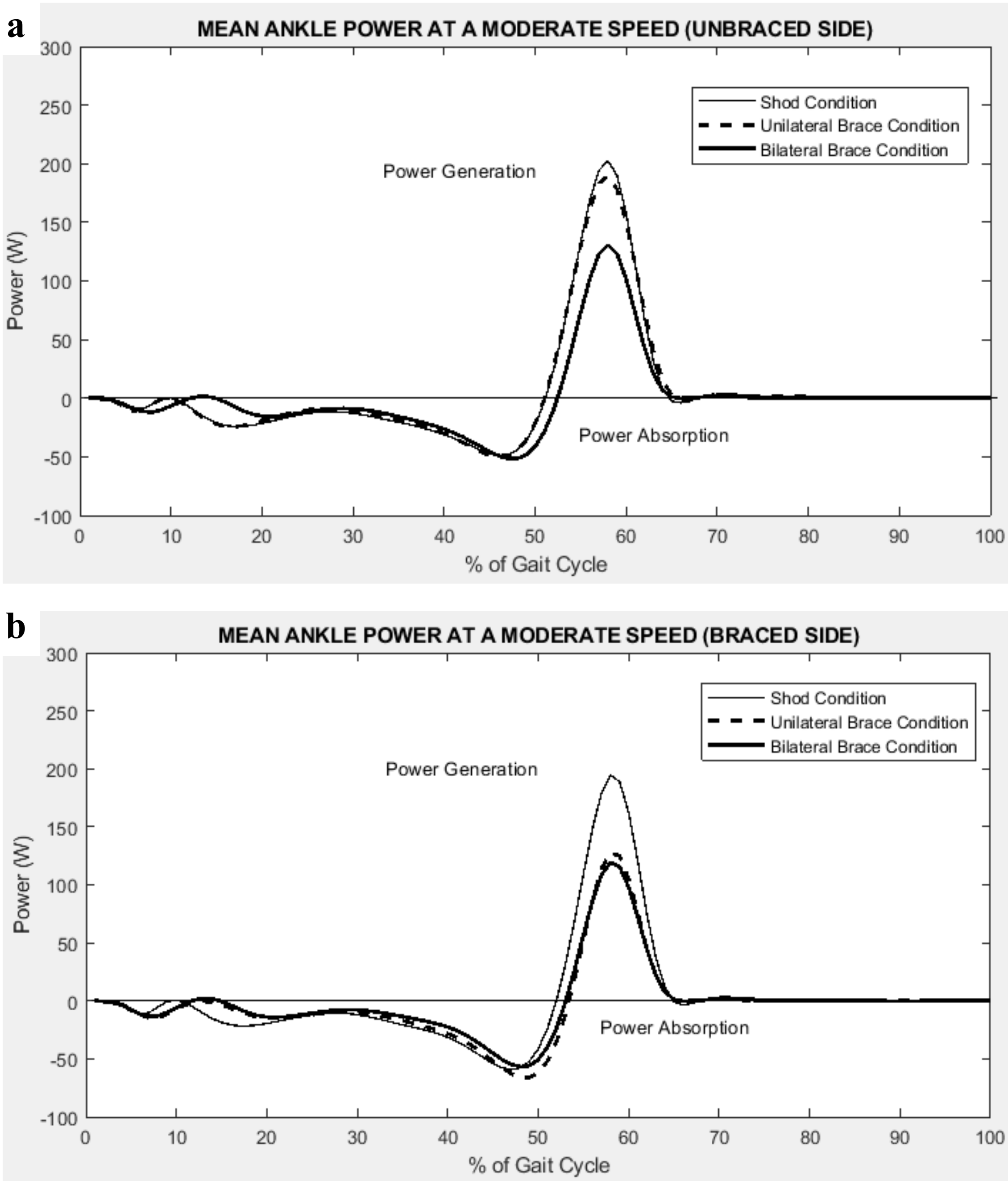


Figure 5*: Mean ankle power at a moderate speed on the (a) unbraced and (b) braced sides.*

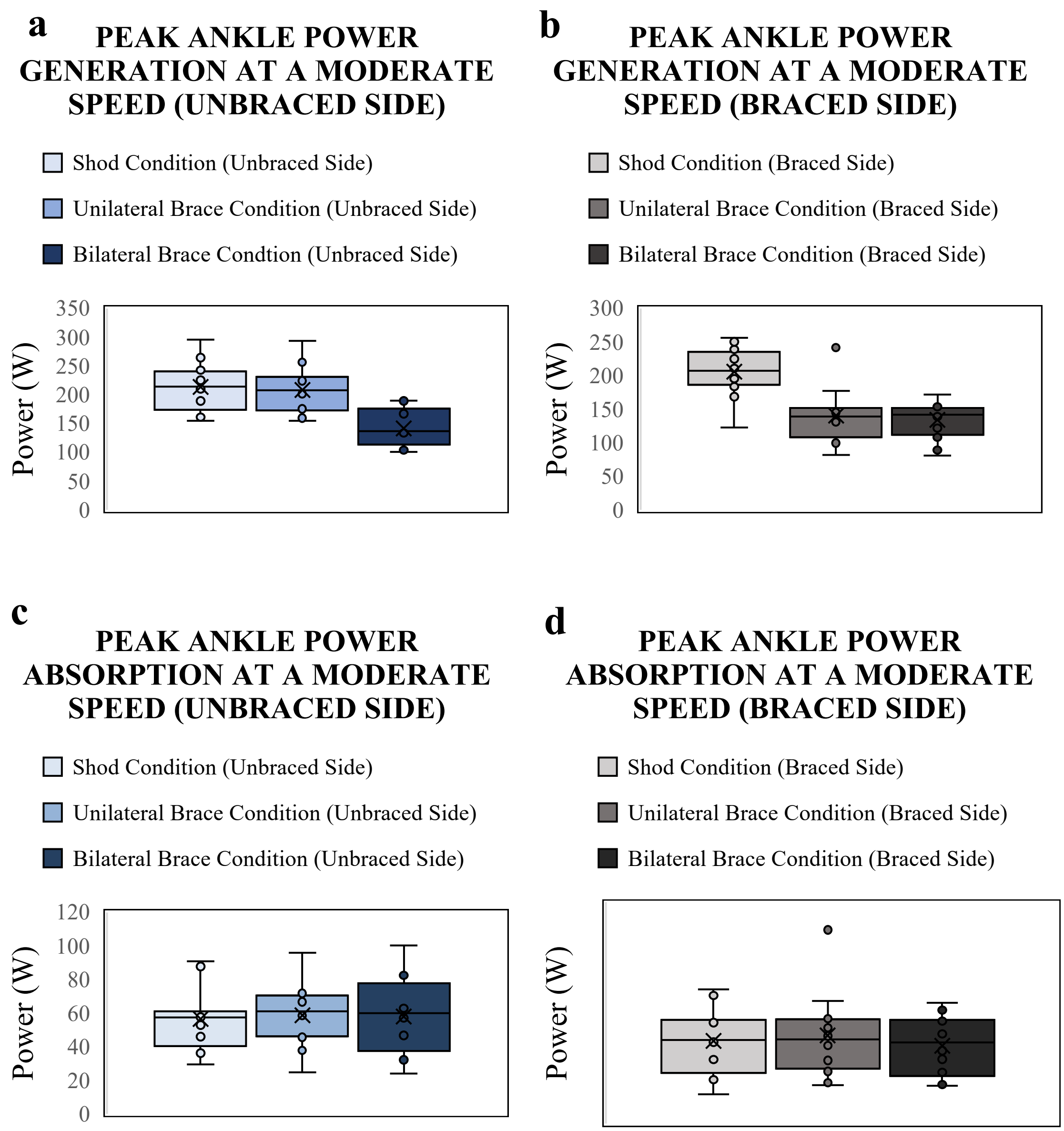


Figure 6*: Peak ankle power generation ((a) unbraced, (b) braced) and absorption ((c) unbraced, (d) braced) at a moderate speed.*

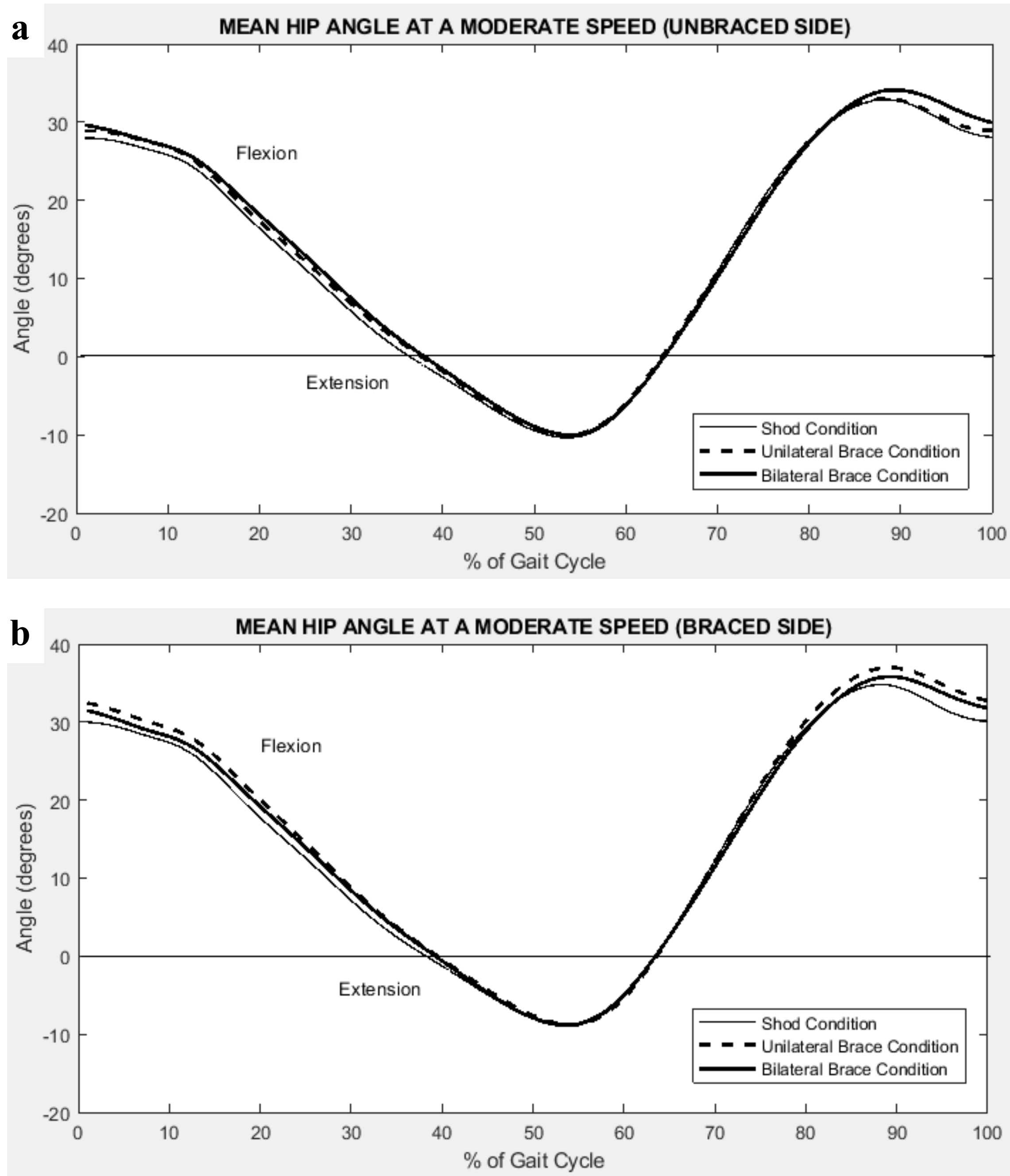


Figure 7*: Mean hip angle at a moderate speed on the (a) unbraced and (b) braced sides.*

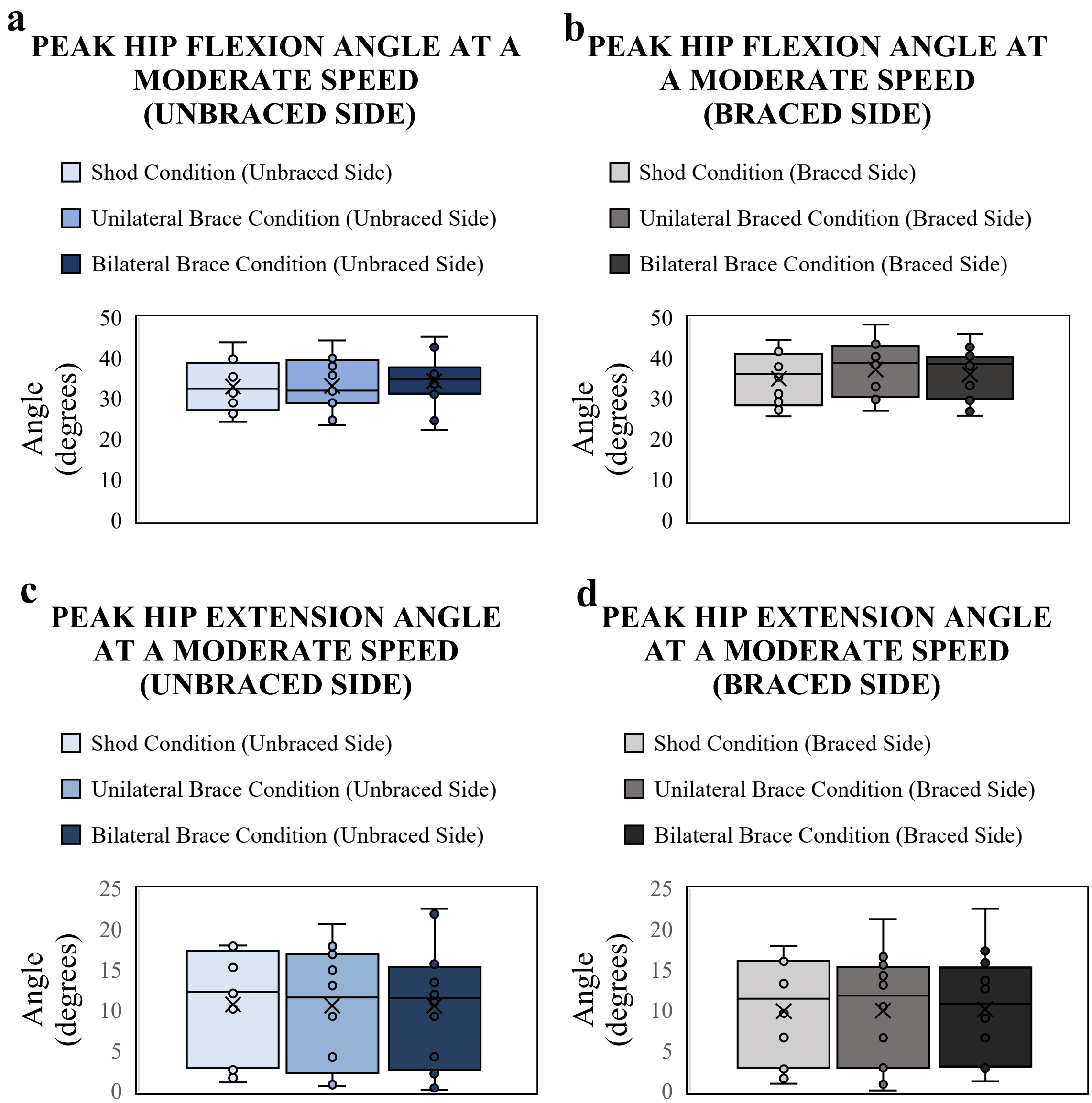
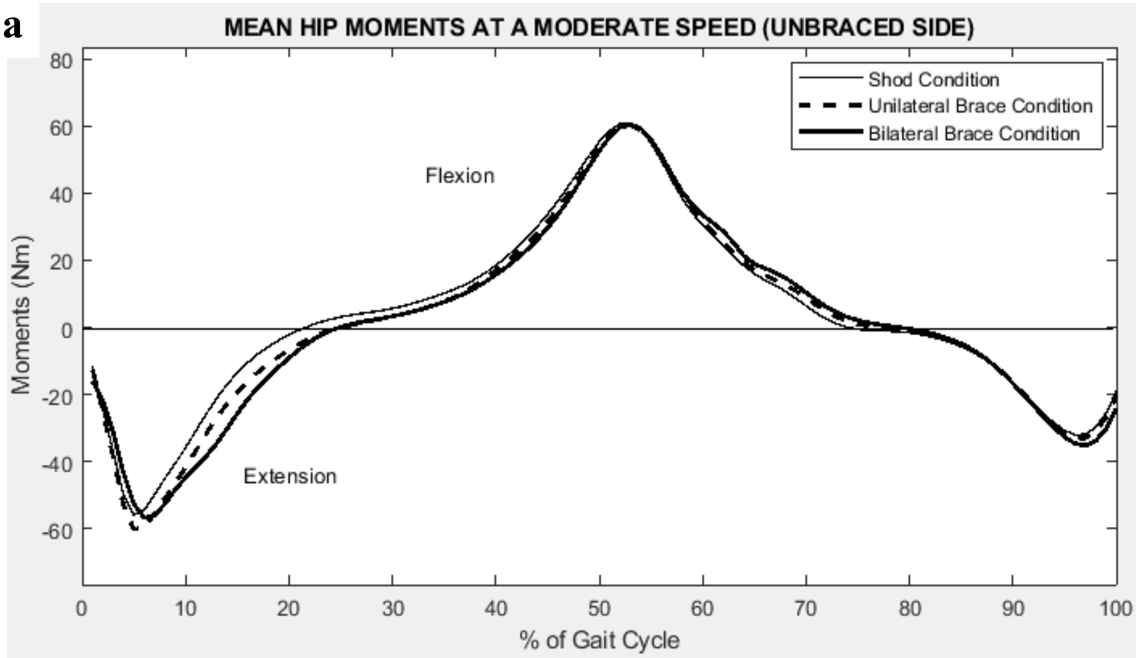


Figure 8*: Peak hip flexion ((a) unbraced, (b) braced) and extension ((c) unbraced, (d) braced) angle at a moderate speed.*

a



b

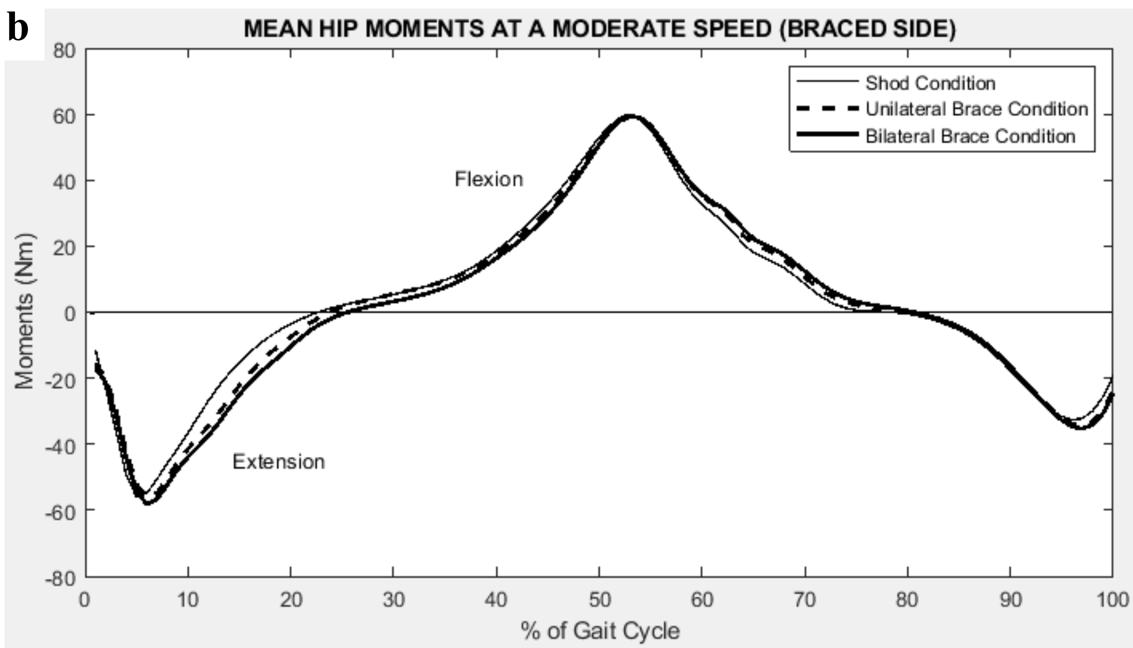


Figure 9*: Mean hip moments at a moderate speed on the (a) unbraced and (b) braced sides.*

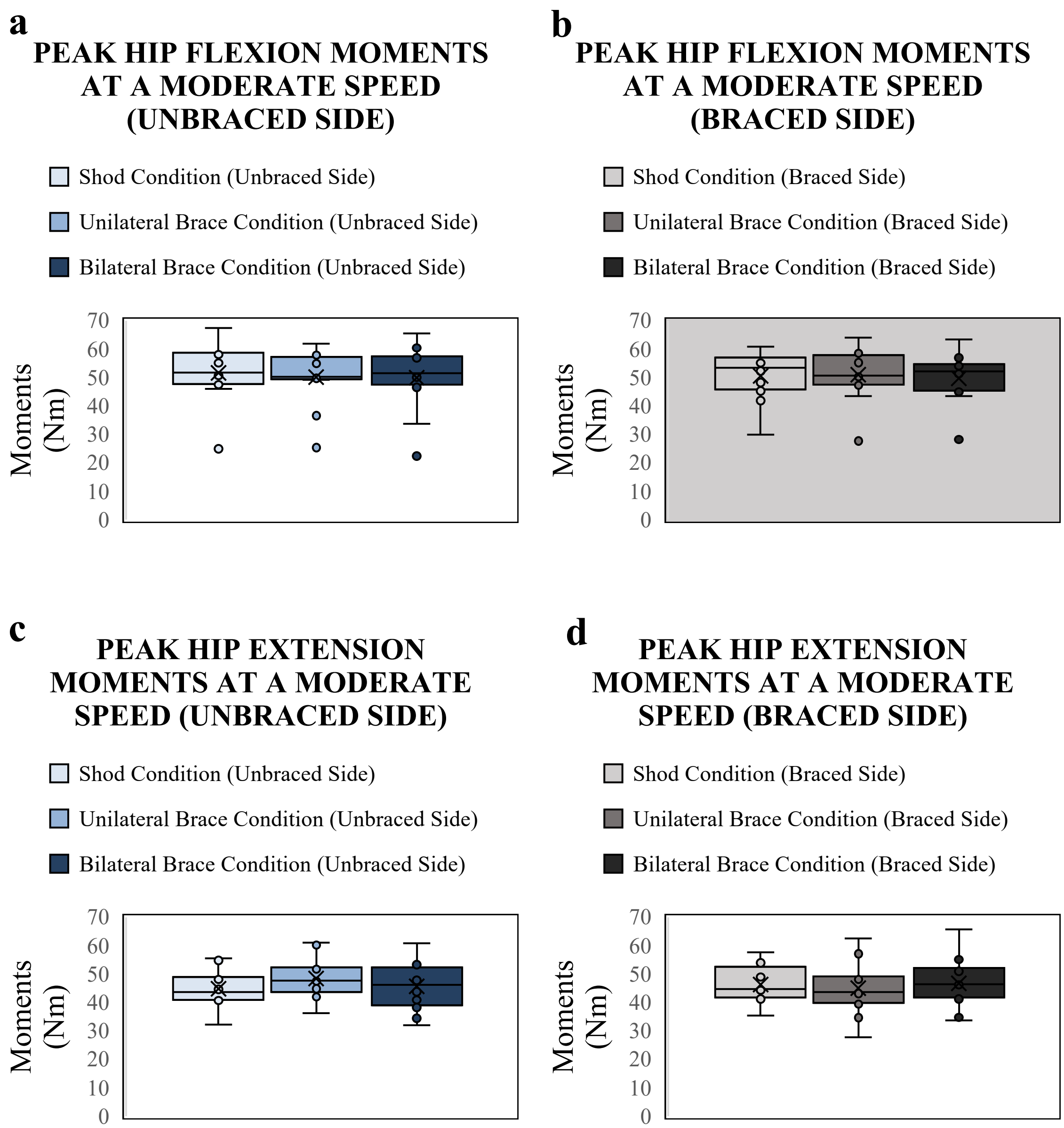


Figure 10*: Peak hip flexion ((a) unbraced, (b) braced) and extension ((c) unbraced, (d) braced) moments at a moderate speed.*

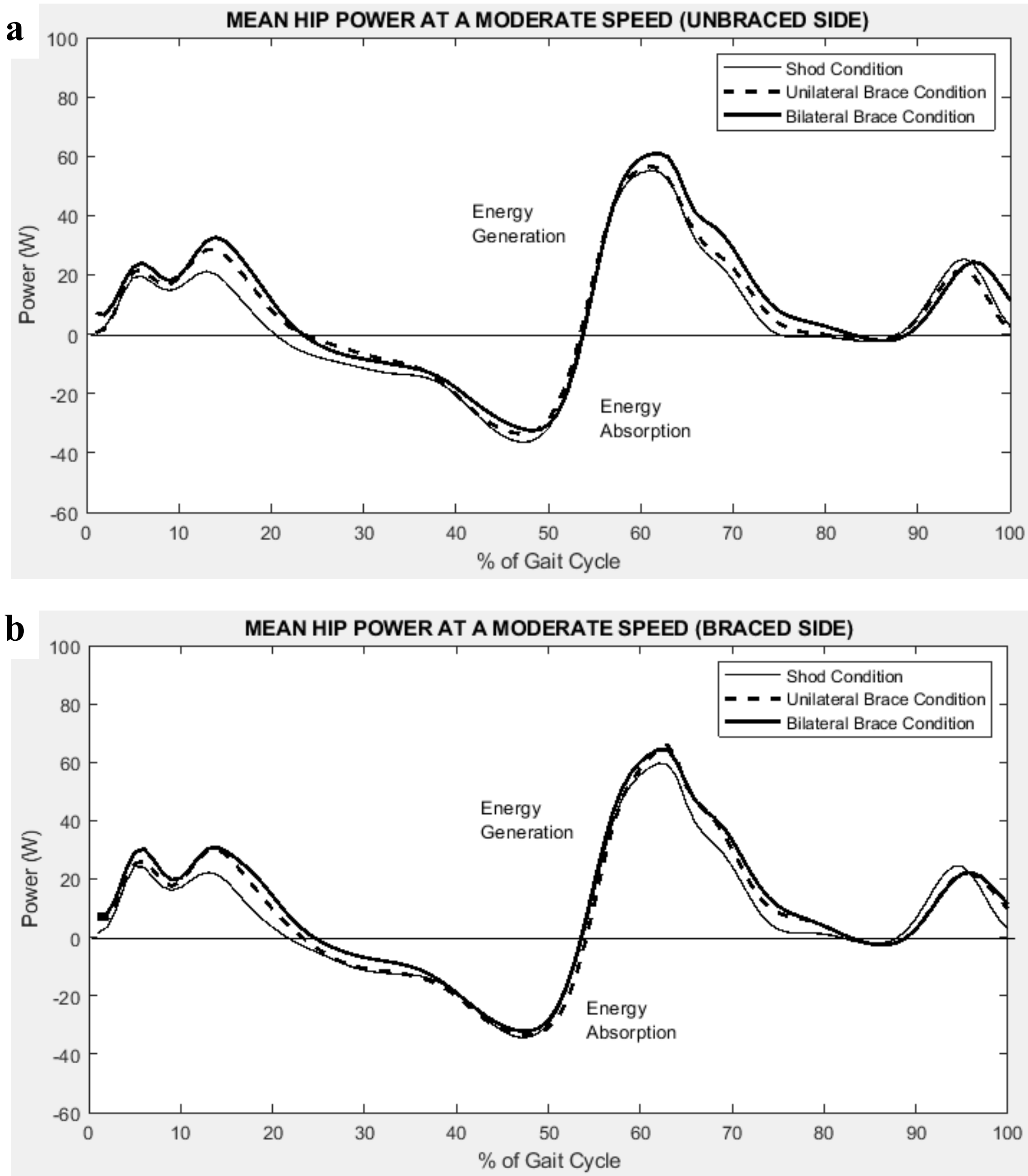


Figure 11*: Mean hip power at a moderate speed on the (a) unbraced and (b) braced sides.*

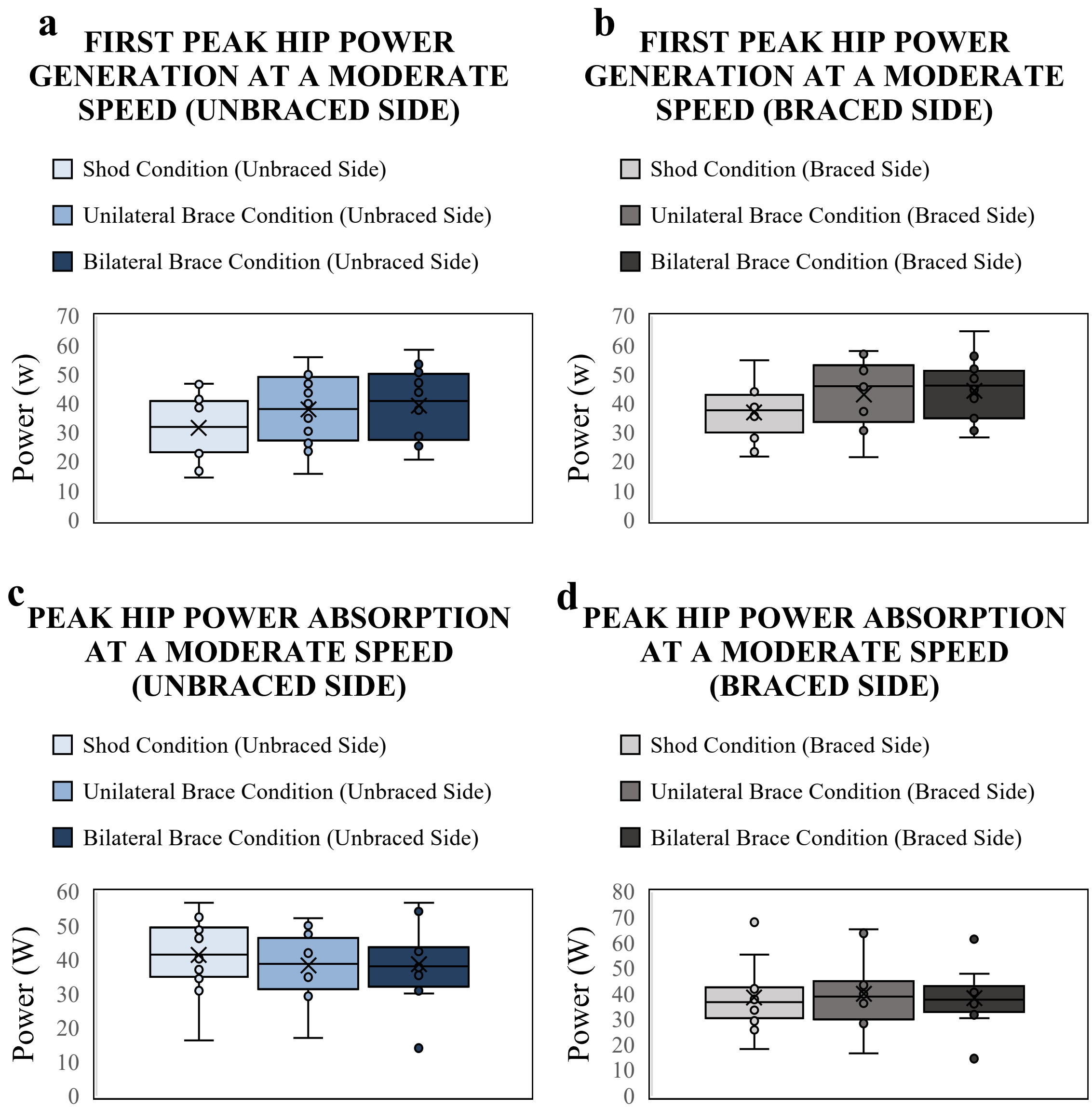


Figure 12: *First peak hip power generation ((a) unbraced, (b) braced) and peak hip power absorption ((c) unbraced, (d) braced) at a moderate speed.*

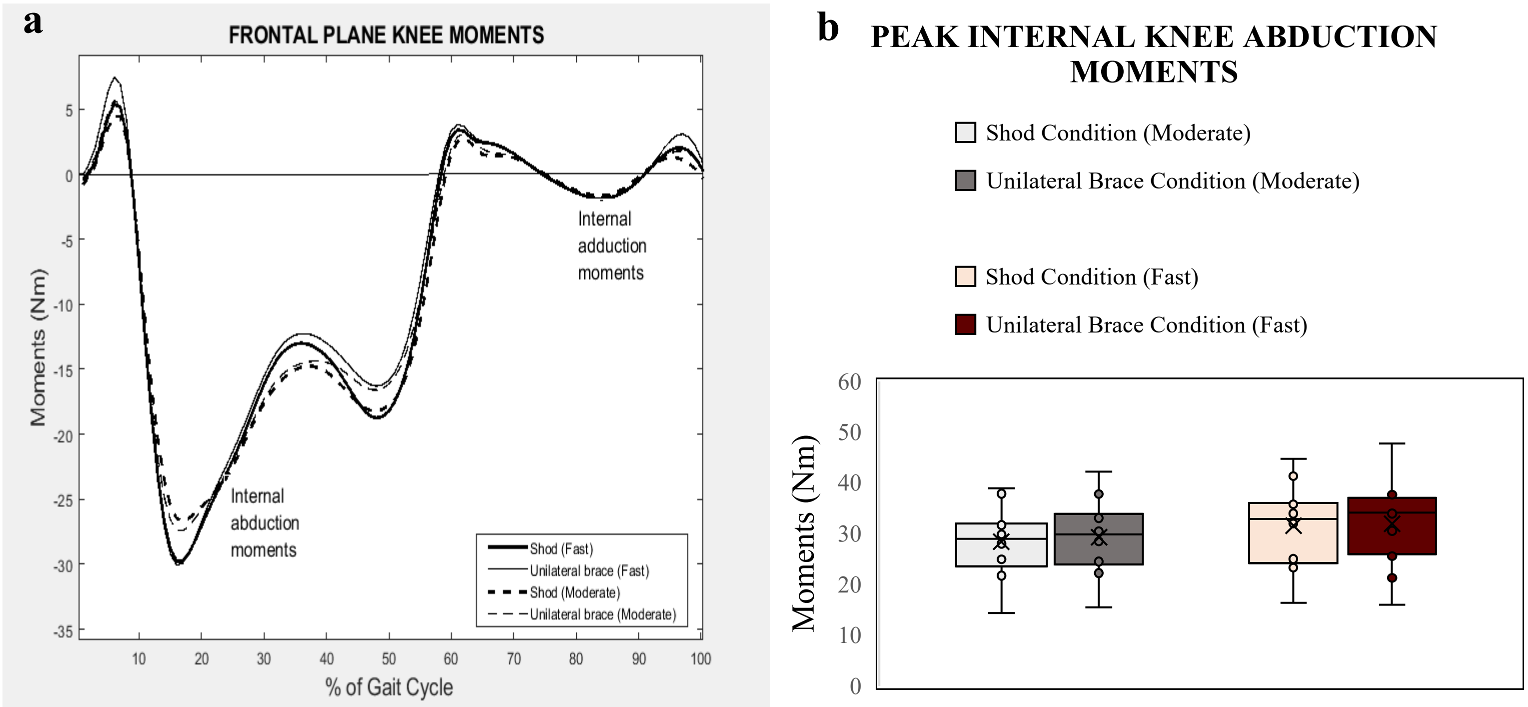


Figure 13: *(a) Mean frontal plane knee moments (b) Peak internal knee abduction moments*

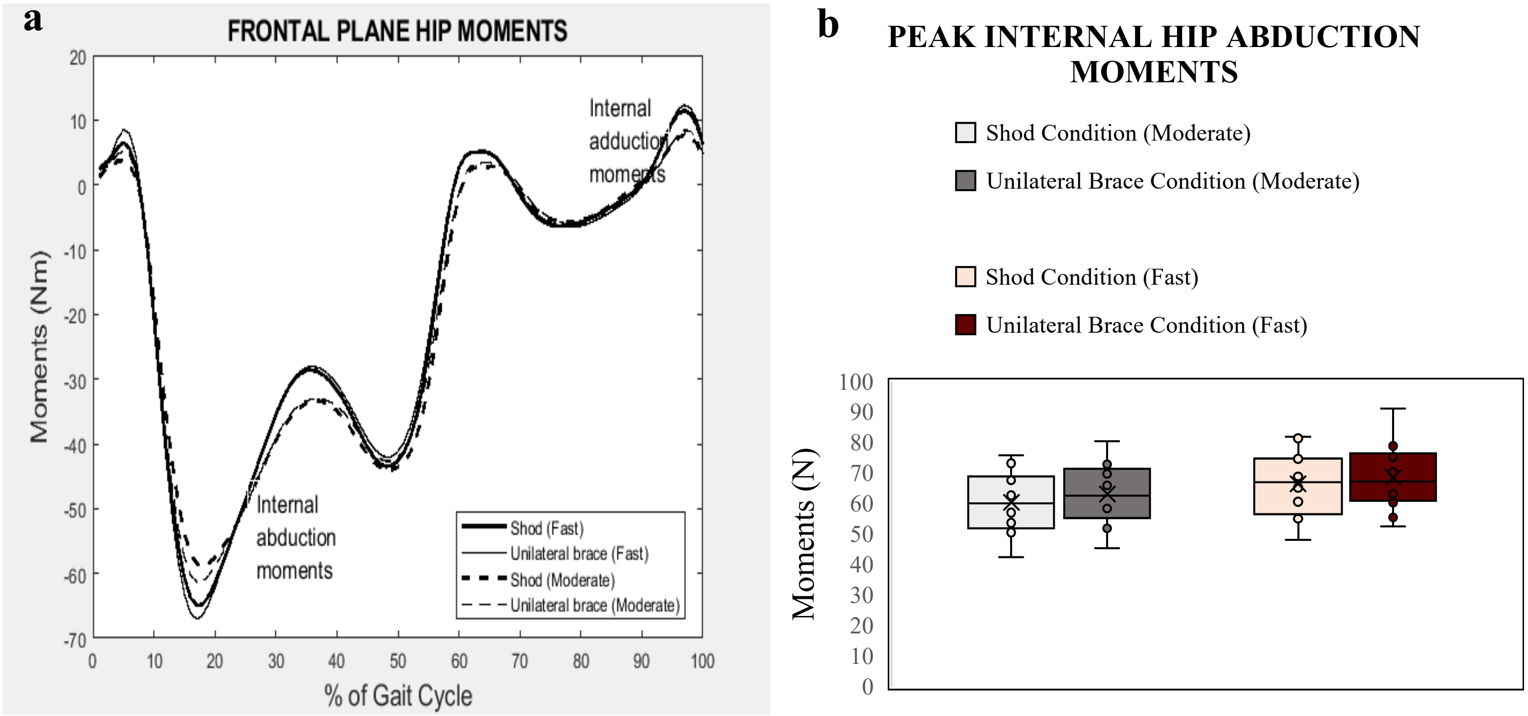


Figure 14: *(a) Mean frontal plane hip moments (b) Peak internal hip abduction moments*

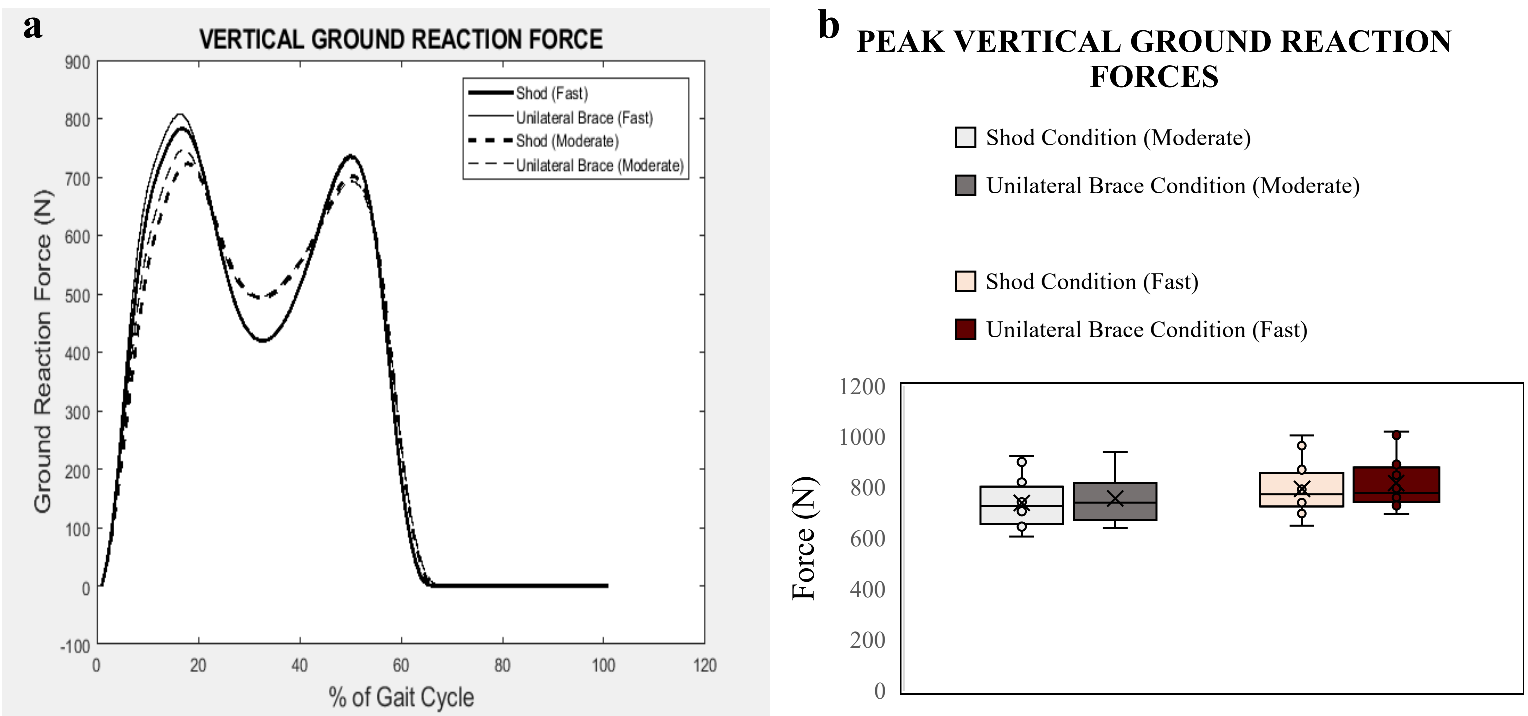


Figure 15: *(a) Mean vertical ground reaction force (b) Peak vertical ground reaction force*

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CURRICULUM VITAE

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