

T.C.
BAHCESEHIR UNIVERSITY
GRADUATE SCHOOL
THE DEPARTMENT OF BIOMEDICAL ENGINEERING

**EFFECT OF A CUSTOMIZED KNEE ANKLE FOOT ORTHOSIS ON JOINT
KINEMATICS AND EMG ACTIVITY ACROSS DIFFERENT SPEEDS**

MASTER'S THESIS
MUSTAFA AL-DHALEMI

ISTANBUL 2025

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ABSTRACT

EFFECT OF A CUSTOMIZED KNEE ANKLE FOOT ORTHOSIS ON JOINT KINEMATICS AND
EMG ACTIVITY ACROSS DIFFERENT SPEEDS

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Biomedical Engineering Master's Program

Thesis supervisor: Assist. Prof. Bora Büyüksaraç

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The biomechanical effect of a customized knee-ankle-foot orthosis (KAFO) on joint kinematics and electromyographic (EMG) activity during gait at different walking speeds is investigated in this thesis. The purpose of the study is to compare gait characteristics between conditions with and without the KAFO to determine the effect of the KAFO on mobility and muscular effort.

Advanced motion capture and EMG systems were used to evaluate twenty participants with respect to parameters including root mean square (RMS), variability (CoV), and range of motion (ROM) across multiple muscle groups and joint movements. Results show that the KAFO reduces variance in muscle activation, increases stability, and changes joint kinematics to enable safer and more efficient gait patterns. Yet, the device imposes constraints on the range of motion at the knee and ankle joints that may affect energy expenditure at higher speeds. The dual role of KAFOs to stabilize lower limbs and to facilitate functional mobility is further emphasized by the findings. The results of this research are critical for clinicians and orthotic designers to understand the importance of designing personalized KAFOs to achieve optimal outcomes. In the end, this work helps to move rehabilitation technologies forward and enhance the quality of life of individuals with lower limb impairments.

Keywords: Gait Analysis, Joint Kinematics, EMG, Lower Limb Biomechanics, Orthotic Design, KAFO

ÖZET

Özelleştirilmiş Diz Ayak Bileği Ortezinin Farklı Hızlarda Eklem Kinematığı ve
EMG Aktivitesi Üzerindeki Etkisi

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Bu tezde, kişiye özel diz-ayak bileği-ayak ortezinin (KAFO) farklı yürüme hızlarında yürüyüş sırasında eklem kinematığı ve elektromiyografik (EMG) aktivitesi üzerindeki biyomekanik etkisi araştırılmıştır. Çalışmanın amacı, KAFO'nun hareketlilik ve kas eforu üzerindeki etkisini belirlemek için KAFO'lu ve KAFO'suz koşullar arasındaki yürüyüş özelliklerini karşılaştırmaktır. Gelişmiş hareket yakalama ve EMG sistemleri, yirmi katılımcıyı birden fazla kas grubu ve eklem hareketi boyunca kök ortalama kare (RMS), değişkenlik (CoV) ve hareket aralığı (ROM) gibi parametreler açısından değerlendirmek için kullanıldı. Sonuçlar, KAFO'nun kas aktivasyonundaki varyansı azalttığını, stabiliteyi artırdığını ve daha güvenli ve daha verimli yürüyüş kalıpları sağlamak için eklem kinematığını değiştirdiğini göstermektedir. Yine de cihaz, daha yüksek hızlarda enerji harcamasını etkileyebilecek diz ve ayak bileği eklemlerindeki hareket aralığına kısıtlamalar getirmektedir. KAFO'ların alt uzuvları sabitlemek ve işlevsel hareketliliği kolaylaştırmak için oynadığı ikili rol, bulgularla daha da vurgulanmaktadır. Bu araştırmanın sonuçları, klinisyenler ve ortez tasarımcıları için, en iyi sonuçları elde etmek için kişiselleştirilmiş KAFO'lar tasarlamanın önemini anlamaları açısından kritik öneme sahiptir. Sonuç olarak, bu çalışma rehabilitasyon teknolojilerinin

ilerlemesine ve alt ekstremitte bozuklukları olan bireylerin yaşam kalitesinin artırılmasına yardımcı olur.

Anahtar Kelimeler: Yürüyüş analizi, eklem kinematığı, EMG, alt ekstremitte biyomekanığı, ortez tasarımı, KAFO



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LIST OF ABBREVIATIONS

KAFO's	Knee Ankle Foot Orthosis
CoV	Coefficient of Variation
MAV	Mean Absolute Value
All RMS	Root Mean Square across all conditions
All peak	Peak value across all conditions
Init contact peak	Peak value at the initial contact phase
Lr peak	Peak value in the loading response phase



Chapter 1

Introduction

1.1 Background on Gait Biomechanics and Assistive Technologies

Human walking or gait biomechanics is the study of walking by looking into how a person walks, muscle coordination, joint coordination, and the way they move such that they can accomplish the task of walking. Gait is repetitive limb movements that produce forward progression in a repetitive sequence using the least amount of energy expenditure possible. The basics of gait include which refers to the movement of body segments as well as the kinetics or forces that move them (Perry & Burnfield, 2010).

While gait analysis first began with the initial visual observation, it has significantly changed following the emergence of the new technology. Sophisticated tools, including modern motion capture systems, and electromyography (EMG) are used today to give a thorough understanding of gait patterns and the biomechanical underpinnings of these patterns. This has allowed for the precise analysis of gait deviations resulting from a variety of pathologies, which are important for rehabilitation and the development of assistive technologies (Baker, 2013).

Knee-ankle-foot orthoses (KAFOs) are commonly used to aid in motivating lower limb-impaired subjects (e.g., those with neuromuscular conditions or traumatic injuries) to maintain better mobility. When an individual with mixed muscular weakness or paralysis uses KAFOs, the mechanical support inside the KAFO stabilizes the knee and ankle joint which leads to a safer, more functional gait pattern (Bapat et al. 2018). Studies have found these devices help compensate for loss of muscular function, and reduce the risk of falls, and enable rehabilitation by facilitating functional gait (Kaufman et al., 1996).

Assistive technologies such as these have become increasingly commonplace in clinical and rehabilitation settings as means of increasing the independence of people with limited mobility. Studies have demonstrated that use of KAFOs can make a profound difference in gait symmetry, stability, and efficiency (Irby et al., 2005), particularly in persons recovering from knee or spinal cord injuries. More importantly,

the emergence of stance-control KAFOs that hold the knee joint in a locked position during the stance phase and unlocked position during the swing phase has improved functional outcomes for users via a more physiological gait pattern (Yakimovich et al., 2009).

1.2 Overview of KAFOs and Their Significance in Mobility Support

Orthotic devices KAFOs support and align the knee, ankle, and foot in order to provide stability and increase functional mobility for those with severe lower limb weakness or paralysis. Different kinds of KAFO include conventional metal and leather KAFO, thermoplastic KAFO, and stance control KAFO. KAFOs are important since they establish a more normal gait, reduce movement, and help improve postural control, improving the QOL of people on mobility assistance (Kaufman et al., 1996). Current conventional KAFOs are typically rigid, offering only constant knee stability thus contributing to an unnatural gait and unnecessary energy expenditure. On the other hand, stance control KAFOs improve gait efficiency because knee flexion is permitted during the swing phase, eliminating the typical movements observed with non-stance control KAFOs (Irby et al., 2005). The type of KAFO that is suitable for a given individual will depend on the level of muscle weakness, amount of joint instability, and personal comfort preference.

1.3 Knee Joint Injuries: Causes, Types, and Common Solutions

Knee joint injuries are responsible for most common injuries, particularly among athletes and older adults. Acute knee injuries stem from trauma, ligament tears or fractures while degenerative conditions like osteoarthritis cause knee injuries. The anterior cruciate ligament (ACL), meniscus, and patella dislocations are the most common types of knee injuries. Certain sports with quick stops and changes in direction are particularly associated with ACL injuries, and meniscal injuries result when the knee is forced to twist inside itself. A degenerative condition, osteoarthritis is more frequent in older adults and results in the slow decomposition of joint cartilage (Griffin et al., 2006).

Treatment of a knee injury generally requires a combination of surgical and nonsurgical approaches. Physical therapy, bracing, and activity modification to take the stress off the knee joint are non-surgical options. KAFOs provide bracing, and the

bracing is very important to stabilize the knee, especially with acute presentations, during the rehabilitation process when we're trying to get a person's mobility back and preserve the integrity of the injured structures. In severe cases, knee stability and function may be restored by ACL repair, a surgical procedure (i.e., reconstruction) (Bapat et al., 2018).

1.4 Injury Rehabilitation and Knee Stabilization and Motivation

The prevention of future complications and recuperation of ability following knee injury are both important. Orthotic devices (e.g. KAFOs) are frequently used during knee stabilization during rehabilitation until the injured structures heal. Utilization of KAFOs in addition aids in preserving the effective joint alignment while allowing equal distribution of weight, thus decreasing the amount of pressure exerted on the affected knee and further permitting ease of rehab (Irby et al., 2005). In order for recovery to be successful, it is highly important to motivate individuals to go along with the rehabilitation protocol. Rehabilitation is a long and difficult process, and it's hard to keep motivated for patients. Patients can use assistive devices to help bring their confidence back to their mobility, leading to their more active participation into rehabilitation programs. Research indicates that people engaged in greater stability feel more secure and are more likely to follow up with their assigned exercises and achieve more functional outcomes (Bapat et al., 2018).

1.5 Research Questions and Hypotheses

The study will address the following research questions:

1. What effect do KAFOs have on lower body kinematics with gait in comparison to individuals who are not KAFO-supported?
2. What are the differences between muscle activation patterns with KAFO-supported and non-KAFO-supported gait?

The hypotheses are:

1. Using KAFO will alter lower body kinematics during gait and improve knee stability.

2. Walking with the KAFO will result in reduced muscle activation patterns, leading to more efficient and natural walking patterns.

1.6 Motivation of the study

The motivation for this study is to improve mobility for persons with lower limb impairments who require the support of a knee-ankle-foot orthosis (KAFO). The main objective is to investigate how the KAFOs change gait biomechanics to allow these individuals to walk more efficiently, with less unnecessary muscle activity, and as close to natural walking patterns as possible. This research seeks to inform clinical practices to optimize orthotic device design and customization based on these effects to improve the quality of life of patients.



Chapter 2

Literature Review

2.1 Overview of Gait Analysis and Its Clinical Applications

Historical Perspective: The evolution of gait analysis has progressed from early observational methods, where clinicians would visually assess a patient's walking pattern, to sophisticated, technology-driven techniques. Early pioneers like Braune and Fischer laid the foundations with basic motion studies, and advancements continued with developments in motion capture systems, electromyography (EMG), and force plates (Baker, 2013). These technologies have significantly enhanced our understanding of the biomechanics of gait, providing quantitative data on joint angles, muscle activity, and forces involved during walking. This transition from qualitative to quantitative analysis has been crucial in enabling personalized treatment strategies for individuals with gait abnormalities.

Modern Clinical Applications: Gait analysis is currently widely used in diagnosing, treating, and rehabilitating gait abnormalities. It is especially significant in assessing functional outcomes of orthotic interventions, such as knee-ankle-foot orthoses (KAFOs). Clinical gait analysis allows for detailed evaluations of how individuals adapt to orthotic devices, providing insights into areas like joint loading, stability, and movement patterns. Such evaluations are critical for optimizing orthotic prescription and rehabilitation programs (Perry & Burnfield, 2010). Moreover, advancements like wearable inertial sensors are making gait analysis more accessible outside laboratory environments, enhancing its application in remote monitoring and personalized rehabilitation (Muro-de-la-Herran et al., 2014).

2.2 Knee Stabilization Methods

Orthotic Interventions: KAFOs are one form of orthotic devices that provide a crucial role in knee stabilization and are used most commonly when individuals have very little or no stability because they are either neurologically or muscularly impaired. KAFOs keep the extremity in the proper alignment, prevent the extremity from moving too far which could result in injury, and allow the patient to ambulate. Recent advances in orthotic design, such as stance control mechanisms that allow free knee motion during the swing phase, have demonstrated the promise of improving gait dynamics

over currently used traditionally locked KAFOs (Yakimovich et al., 2009). Physical Therapy and Muscle Strengthening: Knee stabilization is fundamental in physical therapy including in specific muscle strengthening, stretching (flexibility), and proprioception. Enhancement of stability and support of orthotic use necessitates strengthening exercises for muscles around the knee, namely quadriceps and hamstrings. Other studies have found that combining the use of a KAFO with focused physical therapy leads to the best rehabilitation outcomes including more balanced walking, reduced fall risk, and greater confidence of their walking (Griffin et al., 2006). Orthotic devices provide more stability and proprioceptive training further increases stability such that more efficient movement patterns can be used. Surgical Interventions: For example, surgery, like that of anterior cruciate ligament (ACL) reconstruction is frequently used to stabilize the knee when ligament integrity is compromised. Surgery can restore stability to the anatomy, but following surgery, an orthotic device (such as a KAFO) is used to protect the joint during healing and return the patient gradually back to weight-bearing activities. KAFOs have potential in post-surgical rehabilitation to help with outcomes through reducing stress on healing tissues and allowing for additional stability in physical therapy exercises (Manske & Prohaska 2008).

2.3 The Role and Function of Knee-Ankle-Foot Orthoses (KAFOs) in Gait Rehabilitation

KAFO Types and Importance: The different forms of KAFOs include conventional rigid KAFOs, thermoplastic KAFOs, and more advanced stance control KAFOs. Rigid conventional KAFOs maximize stability, but at the expense of the natural motion of the knee thereby potentially increasing the energy cost of gait. A lighter alternative, thermoplastic KAFOs improves user comfort and compliance. A finite number of studies have demonstrated that the use of stance control KAFOs, which permit knee bending during the swing phase and stability during stance, results in improved gait economy and reduction in hip movements leading to a more natural walk (Yakimovich et al., 2009). **KAFOs in Neuromuscular Conditions:** KAFOs are essential aids in maintaining stability and improving functional mobility for individuals with neuromuscular impairments, such as spinal cord injury, polio, and

muscular dystrophy. KAFOs have been shown through research to improve the ability to walk, reduce fatigue, and allow daily functional activity and independence, (Brehm et al., 2008). Furthermore, KAFOs enable the reduction of the risk of joint contractures and deformities by sustaining appropriate position, which is chief during patients with muscle weakness and spasticity.

2.4 Biomechanical Implications of KAFO Use

Gait Parameters Affected by KAFO Use: KAFO use greatly modifies joint kinematics, muscle activation patterns, and energy expenditure during gait. Kaufman et al. (1996) have studied through the use of motion analysis systems and concluded that KAFOs serve as stabilizers of knee and ankle joints resulting in more consistent stride lengths and reduced gait asymmetry. However, some users may hike or circumvent the hip more than normal to clear the orthotic limb during the swing phase.

Kinematics and EMG (KAFO and Gait): The first is that gait kinematics of KAFO-supported gait have less knee flexion during the swing phase. Although there are no measures taken of motion pertaining to the knee joints, they relate to studies using electromyography (EMG) in which there were variations in the muscle activation patterns, specifically heightened use of proximal muscles (gluteus medius) for reduced knee mobility (Irby et al., 2005). These results emphasize the need for a more detailed understanding of individual muscle recruitment strategies for designing improved KAFOs that may enhance gait efficiency. **Energy Efficiency and Muscle Activation:** For a patient walking with KAFOs, the energy cost is usually higher than when walking unassisted, and especially so with conventional locked KAFOs. Seated PLA velocity profiles provide the 031, but not the 221 muscle ### Explanation: Lower energy expenditure and lower muscle activation can lead to less muscle compensation, which, in turn, facilitates a more efficient gait pattern (Brehm et al., 2008). Long-term use of the KAFO and improving the quality of life of KAFO users requires reducing energy costs.

2.5 Comparative Studies Involving KAFOs versus Other Orthotic Aids or Unassisted Conditions

KAFOs vs. No Orthotic Aid: Comparative studies highlight that individuals using KAFOs exhibit improved gait stability and reduced risk of falls compared to those without orthotic support. These benefits are particularly evident in populations with lower limb weakness, where KAFOs provide the necessary joint stability for safe ambulation (Irby et al., 2005). Additionally, KAFOs contribute to a reduction in strategies, such as excessive trunk lean or hip hiking, which are often observed in unassisted gait.

KAFOs vs. Other Orthotic Devices: Comparisons between KAFOs and ankle-foot orthoses (AFOs) show that while AFOs are effective for stabilizing the ankle and supporting foot drop, KAFOs offer superior knee stabilization, making them more suitable for individuals with combined knee and ankle instability (Yakimovich et al., 2009). However, the increased weight and bulkiness of KAFOs compared to AFOs can impact user comfort and compliance, emphasizing the need for individualized orthotic prescriptions.

2.6 KAFO Use in Rehabilitation and Functional Outcomes

Rehabilitation Outcomes: For individuals with a history of knee injury involving ACL tears or as part of the rehabilitation program following total knee arthroplasty, KAFOs are integrated into the treatment program. During rehabilitation exercises, these orthoses stabilize patients, so that they can perform weight-bearing without additional injury. The use of KAFOs during the early stages of rehabilitation has been shown in research to improve functional outcomes, by setting appropriate joint alignment and protecting healing tissues (Griffin et al., 2006). Additionally, KAFOs allow for an escalation of activity levels as bone, muscle, and joint proprioception return to normal.

Impact on Quality of Life: KAFOs impact the quality of life in the lives of individuals in various capacities by increasing mobility, independence, and confidence. In fact, studies on patient-reported outcomes suggest that while patients using KAFOs are less satisfied with their ability to walk compared to cohorts using other types of devices, those using KAFOs do rate higher satisfaction for walking in

the community setting where uneven terrain presents difficulty (Brehm et al. 2008). In addition to the physical benefits, the use of KAFO reduces fear of falling and allows the patient to participate more in social activities which contributes to better well-being.

2.7 Filling the Gap in Current Literature

Limited Studies on Long-Term Use: Though documented benefits of KAFOs exist, there is a lack of long-term studies analyzing effectiveness, compliance, and user satisfaction over a long period of time. Much of the current research is focused on short-term outcomes and raises questions regarding the sustainability of gains and adverse events associated with extended use (Ferris et al., 2012). However, long-term studies are necessary to learn how factors such as device wear, variations in physical condition, and evolving user needs influence KAFO effectiveness.

Variability in KAFO Designs and User Needs: Current research often overlooks the variability in KAFO design and the requirements for bootmaker customization. When studying KAFOs, findings are often generalized across different types of KAFO without considering differences in patient's anatomy, gait mechanics, and functional goals. To fill these gaps a more personalized approach would be required, including fitting to meet individual needs and adjusting appropriately to optimize gait outcomes and increase user satisfaction (Owen, 2010).

2.8 Rationale for the Comparative Approach in the Study

Need for Comparative Data: A comparative study will be conducted on KAFO-supported versus non-KAFO-supported gait in order to bridge existing knowledge gaps. We therefore discuss progress toward the development of better clinical decision-making informed by how KAFOs alter kinematic parameters and muscle activation compared to unassisted conditions. These data are required for the formulation of evidence-based guidelines for orthotic prescription and rehabilitation strategies.

Potential Contributions to Clinical Practice: These comparative analysis findings may have pronounced consequences in clinical practice for optimal KAFO design and rehabilitation protocols. The biomechanical and functional influence of KAFO use can

be better understood and this can inform clinician's selection of the best orthotic intervention to improve patients' outcomes and quality of life.



Chapter 3

Methodology

3.1 Custom-Made KAFO Knee Brace Design

This study uses a custom-made knee brace Knee-Ankle-Foot Orthosis (KAFO) which utilizes a spring-loaded mechanism to increase knee bending flexibility. This new design intends to minimize the contact pressure of the knee joints while supporting dynamic movement. The design includes a rigid spring mechanism placed posteriorly which provides optimal support during flexion and extension. Maximizing comfort during extended use, the brace is fabricated using lightweight, breathable materials. This integrated spring system produces resistance and flexibility over a limited range of motion best suited for walking speeds from slow to moderate.



Figure 1. KAFO brace design

3.1.1 Biomechanical analysis. Biomechanical analysis techniques, including joint loading, were employed to evaluate the effectiveness of the custom KAFO. This method is compatible with the methodology proposed by Yakimovich, Lemaire, and

Kofman (2009) who underline the value of biomechanical analyses for assessment of performance of orthotic devices (Yakimovich et al, 2009).

3.1.2 Spring-Loaded Mechanism. The inspiration for a spring-loaded element of this custom KAFO incorporated the designs evaluated by Yakimovich, Lemaire, and Kofman (2009). Through these mechanisms joint movement and unloading are dynamically assisted, an approach that has already proven encouraging in decreasing joint stress during movement (Yakimovich et al., 2009).



Figure 2. Spring loaded mechanism.

3.1.3 Materials and Manufacturing. Recent advancements in additive manufacturing and material science allowed the team to select lightweight yet durable materials to use in the selection, guided by studies such as those by Khajavi and colleagues (2023). Khajavi et al. (2023) stress the use of advanced engineering materials in raising device efficiency and providing the user a comfort consideration for device operation.

3.2 Participants

A total of 20 healthy adult participants (mean age: 24. This study recruited subjects (age: 28 ± 3.8 years (mean $\pm$ 1SD); height range: 1.760 – 1.80 m (mean height: 1.75 ± 0.1 m); and weight range: 70 – 100 kg (mean weight: 70.5 ± 9.2 kg). No participants had a history of lower limb injury or mobility issues. Prior to recruitment ethical approval was obtained and all participants signed informed consent forms. To simulate real-world walking conditions, participants were required to walk on a motorized treadmill at three distinct speeds: 2 km/h, 4 km/h, and 6 km/h.

3.3 Experimental Setup

3.3.1 Devices Used.

1. Surface Electromyography (sEMG): Muscle activation was recorded using the Delsys Trigno system. This allowed for the measuring of precise data from this system consisting of 14 wireless sensors sampling at a frequency of up to 2000Hz. Two wireless sensors were placed on the sensor sites as per SENIAM guidelines.



Figure 3. Delsys trigno system.

2. Motion Capture System: Kinematic analysis was performed using an Xsens Awinda IMU-based system. To track lower body motion, the system includes 7 IMUs located strategically on key anatomical landmarks. The data were gathered at a frequency of 60 Hz.



Figure 4. Inertial measurement units (IMU).

3. Treadmill: Walking speeds were standardized by a motorized treadmill.



Figure 5. Motorized Ultima treadmill.

4. Synchronization Module: EMG and motion capture data were acquired simultaneously while timed to synchronize with the Delsys trigger module.



Figure 6. Delsys' trigger module.

3.3.2 Muscle Groups Studied. The study focused on the following muscles due to their significant roles in locomotion:

1. Biceps Femoris

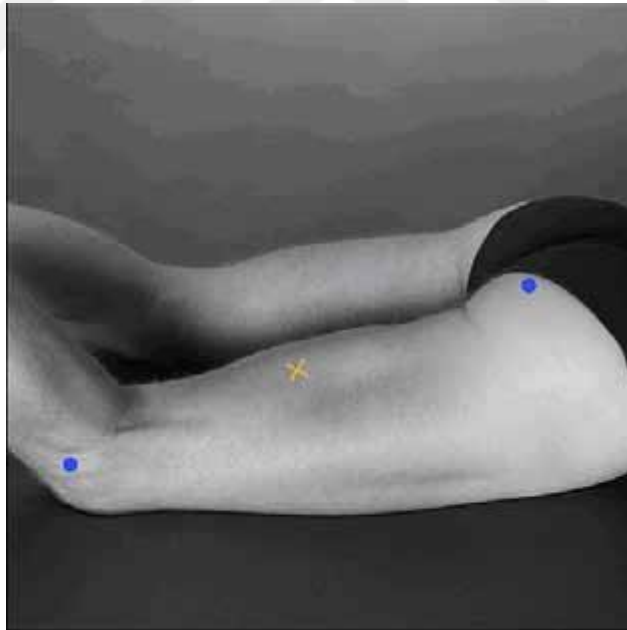


Figure 7. Position of EMG electrode (SENIAM) for Biceps Femoris

2. Gastrocnemius Lateralis

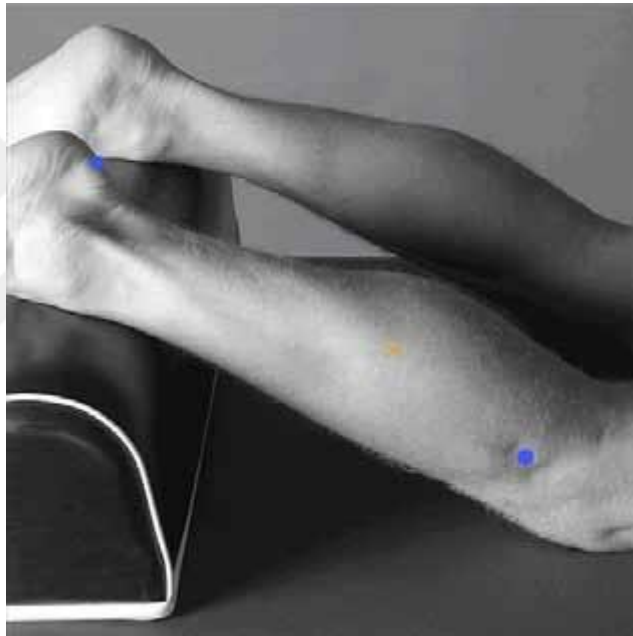


Figure 8. Position of EMG electrode (SENIAM) for Gastrocnemius Lateralis

3. Gastrocnemius Medialis



Figure 9. Position of EMG electrode (SENIAM) for Gastrocnemius Medialis

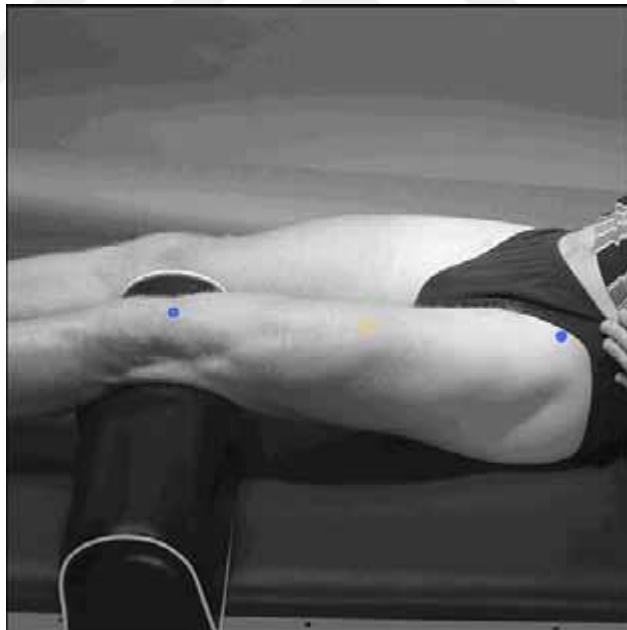


Figure 10. Position of EMG electrode (SENIAM) for Rectus Femoris

4. Tibialis Anterior



Figure 11. Position of EMG electrode (SENIAM) for Tibialis Anterior

5. Vastus Lateralis



Figure 12. Position of EMG electrode (SENIAM) for Vastus Lateralis

6. Vastus Medialis

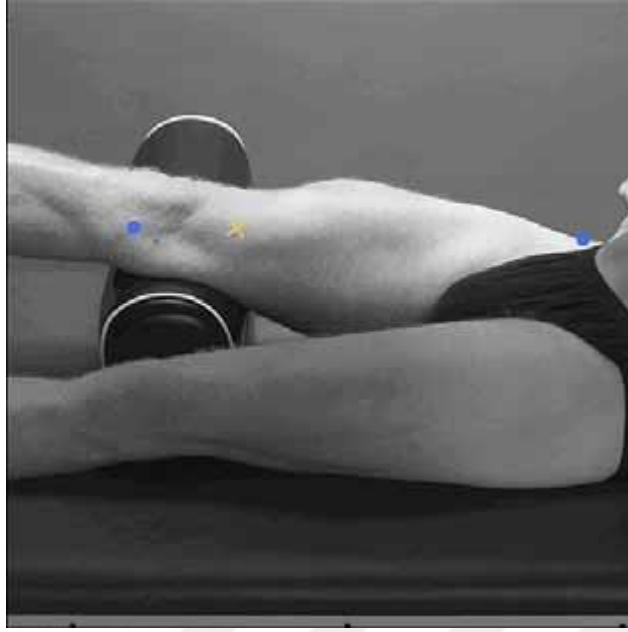


Figure 13. Position of EMG electrode (SENIAM) for Vastus Medialis.

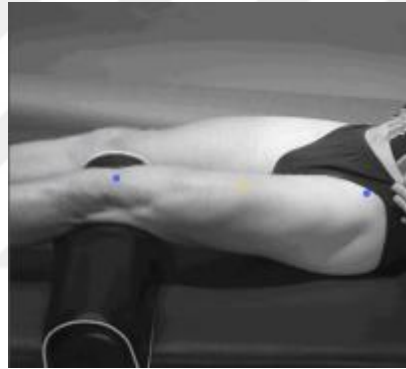


Figure 14. Position of EMG electrode (SENIAM) for Rectus Femoris

3.4 Data Collection Protocol

3.4.1 Participant Preparation. Lightweight athletic clothing was worn to minimize impedance to sensor placement. To achieve proper electrode adhesion on specific muscle landmarks and reduce EMG signal noise, muscle landmarks were cleaned. The Xsens motion capture system was calibrated using anthropometric measurements.

3.4.2 Data Acquisition. The walking trials (with and without KAFO brace) were repeated two times for each participant at each speed. To reduce the potential bias, trials were randomized. To avoid fatigue, Participants sat quietly for 5 minutes

between trials. During each 2-minute trial, EMG and motion data were simultaneously recorded.

3.4.3 Sensor Placement. Seven IMU sensors were placed on key anatomical landmarks: legs (upper legs, lower legs, and feet), and pelvis. Seven EMG sensors for each leg were positioned to monitor muscle activity in the rectus femoris, biceps femoris, Gastrocnemius Medialis, Gastrocnemius Lateralis, Tibialis Anterior, Vastus Lateralis and Vastus Medialis. Gastrocnemius Medialis, Gastrocnemius Lateralis, Tibialis Anterior, Vastus Lateralis and Vastus Medialis.



Figure 15. Position of XSENS sensor for the lower body part (Pelvis, Upper Leg, Lower Leg, Foot).



Figure 16. The XSENS sensors on the lower body part were used during the experiment.

3.5 Data Processing

3.5.1 EMG Data Processing. EMG signals were processed in MATLAB using the following steps:

1. **Noise Filtering:** A 4th-order Butterworth bandpass filter (10-250 Hz by default) was applied to remove low-frequency drift and high-frequency noise, ensuring signal clarity.
2. **Rectification and Smoothing:** EMG signals were rectified (absolute value taken) to retain signal magnitude. EMG is smoothed with optional technical such as a moving average window or low pass filtering, to generate an EMG envelope.
3. **Gait cycle detection:** Heel strike markers marked by the XSENS motion capture system were used to segment the data into individual gait cycles. EMG signals of each gait cycle were normalized to 100% of the cycle for consistent comparison of variabilities across strides. RMS (Root Mean Square) and peak values were calculated for each gait cycle to determine muscle activity.
4. **Sensor landmarks** were cleaned and prepared to ensure proper electrode adhesion and reduce noise in the EMG signals.

3.5.2 Processing Motion Capture Data. The kinematic parameters were automatically extracted from Xsens motion capture data including segments' positions and lower limb joint angles. Heel-strike and toe-off events were used to identify gait cycles and to align with EMG data with precision. Flexion/extension, abduction/adduction, internal/external rotation were analyzed in terms of the range of motion (ROM) and mean/max angles at the joints. An anthropometric assessment was conducted to calibrate the Xsens motion capture system.

3.6 Parameters Calculated

3.6.1 EMG Parameters. The following parameters were calculated to comprehensively analyze muscle activity:

RMS: Average root mean square of muscle activation for all cycles. If the conditions under the brace are associated with higher RMS this indicates either that the muscle has increased its force output, perhaps to account for changes in joint

dynamics, or simply that the muscle is engaging more. In either case, the higher RMS is suggestive of higher CoP, but that change in the instability metrics has likely occurred because of changes to other aspects of the task. Reduced effort may be indicated by lesser RMS as a result of brace support.

$$\text{RMS} = \sqrt{\frac{1}{N} \sum_{i=1}^N x_i^2} \quad (1)$$

Where:

- N = Total number of data points.
- x_i = Value of the signal (e.g., EMG amplitude) at the i point.
- $\sum_{i=1}^N x_i^2$ = Sum of the squared values of the signal.

PEAK: Peak value reached through maximum across gait cycles. This may suggest that peaks are higher with the brace, and these may represent bursts of activation. Smother, or lesser, activation may mean lower peaks.

$$\text{Peak} = \text{MAX}_{i=n} (x_i) \quad (2)$$

- x_i = The value at each data point i .
- n = Total number of data points in the measurement period
- max = The maximum function, selecting the largest value in the dataset.

INITIAL CONTACT PEAK: A maximum muscle activation at the toe off. Stabilization muscles at initial contact may be more relied on with elevated peaks with the brace. Smother muscle recruitment is associated with lower peaks.

$$\text{Initial Contact Peak} = \text{MAX}(x(t)) \quad (3)$$

max = The maximum function to find the peak value in the defined window.

$x(t)$ = The value of the variable of interest (e.g., EMG) at time t .

LOADING RESPONSE PEAK: The peak muscle activation was seen at the loading response phase. If higher peaks, then one's stabilization needs may be greater. If the load is reduced, should result in lower peaks from the brace.

$$\text{Loading Response Peak} = \max (x(t)) \quad (4)$$

- $x(t)$ = The variable of interest (EMG) at time t_i .
- $\max$ = The maximum function to identify the peak value within the phase.

Coefficient of Variation (CoV): The variability, expressed as the ratio of standard deviation to mean. Increased variability in activation of these muscles is further suggested by higher CoV with the brace. More consistent recruitment patterns have lower CoV.

$$CV = \left(\frac{\sigma}{\mu} \right) \times 100 \quad (5)$$

σ = Standard deviation of the dataset.

μ = Mean (average) of the dataset.

CV is expressed as a percentage (%).

Mean Absolute Value (MAV): The EMG signal absolute values mean. Greater muscle effort is associated with a higher MAV with the brace. Reduced effort due to mechanical support from the brace is characterized by lower MAV.

$$MAV = \frac{1}{N} \sum_{i=1}^N |x_i| \quad (6)$$

4.6.2 Kinematic Parameters

Joint Range of Motion (ROM): The ROM of a joint measures the size of the movement of a joint throughout the gait cycle from the difference between the maximum and minimum joint angles. It is used to assess the flexibility and functional range of the joint, which is crucial for understanding normal versus pathological gait patterns.

$$ROM = MAX - MIN \quad (7)$$

Joint Root Mean Square (RMS): RMS evaluates the variability and consistency of joint angles over the gait cycle. It provides a measure of movement stability and coordination. Its role is to analyze how well joints maintain stable and controlled movements during dynamic tasks such as walking.

$$RMS = \sqrt{\frac{\sum_{i=1}^n (\theta_i - \underline{\theta})^2}{n}} \quad (8)$$

Where θ_i the joint angle at the time i , $\underline{\theta}$ is the mean joint angle, and n is the total number of data points in a gait cycle.

Maximum Joint Angle (Max): The highest joint angle value observed during the gait cycle. Used to evaluate the peak range of movement, which reflects the joint's ability to extend or flex fully.

$$\theta_{max} = (\theta_1, \theta_2, \dots, \theta_N) \quad (9)$$

θ_i is the angle at time i .

Minimum Joint Angle (Min): The lowest joint angle value recorded during the gait cycle. Used to establish the baseline joint position and detect limitations in movement toward a neutral or resting position. Important for detecting restrictions or structural impairments that limit joint extension or flexion. Essential in comparing minimum angles across assisted and unassisted walking conditions.

$$\theta_{min} = (\theta_1, \theta_2, \dots, \theta_N) \quad (10)$$

θ_i is the joint angle at the time i .

3.7 Statistical Analysis

All data were tested for normality using the Kolmogorov-Smirnov test. Depending on the nature of the distribution, the parameters comparisons between the with and without brace conditions were done using the Independent samples T-Test (if normal distribution was assumed) or its Non-parametric alternative the Wilcoxon rank-sum test (for non-normally distributed data). Statistical significance was set at $p < 0.05$.

Chapter 4

Findings

The parameters are calculated for a comprehensive understanding of muscle activity and joint kinematics during gait, and to explore the role of knee braces. The data collected in this chapter is presented in tables, stipulated according to the parameters previously selected. For clarity and ease of interpretation, the results are categorized into two primary data types: muscle activity and joint kinematics.

4.1 Muscle activity results

The previously mentioned parameters were selected to represent muscle activation levels, variability, and timing, which are important for gauging the effects of wearing a knee brace during gait at various speeds (2 KMH, 4 KMH, and 6 KMH).

Results are presented in the following tables in terms of median values, as data were not normally distributed. Medians are used for robust representation of central tendencies in the presence of non-normality.

This study selected parameters that offered insight into muscle engagement, strategies, and biomechanical adaptations caused by the knee brace. The following sections present and discuss these parameters in detail.

4.1.1 RMS of muscle activation results. Table 1 shows that the Biceps Femoris muscle had increased activation with the knee brace at all walking speeds. The RMS value increased by 0.074 mV at 2Km/H with brace compared to the unbraced ($p < 0.05$) condition. The increase was more moderate at 4Km/H (0.030 mV) ($p < 0.05$) and even more so at 6Km/H (0.026 mV) ($p < 0.05$). The results show that the knee brace results in slightly higher muscle activation in the Biceps Femoris, but the effect decreases with increasing speed.

In general, the RMS values decreased with the brace for the Gastrocnemius Lateralis. A reduction of -0.080 mV was observed at 2 km/H ($p < 0.05$). The difference was minimal at 4Km/H ($p = 0.82$) (0.002 mV) and at 6Km/H, the RMS decreased by -0.047 mV ($p < 0.05$). These results suggest that the knee brace decreases activation of the Gastrocnemius Lateralis, especially at slower walking speeds.

The Gastrocnemius Medialis showed minimal differences between the braced and unbraced conditions at all speeds. For 2Km/H the difference was -0.005 mV ($p =$

0.47), 4Km/H was 0.001 mV ($p = 0.33$) and 6Km/H was -0.020 mV ($p < 0.05$). The results of these findings indicate that the knee brace has little or no effect on the activation of this muscle.

At higher speeds, the Rectus Femoris showed slight reductions in activation with the brace. The RMS difference was minimal at 2Km/H (-0.009 mV) ($p = 0.35$). The larger reduction (-0.027 mV) ($p = 0.05$) was caused by the brace at 4Km/H and at 6Km/H the decrease was -0.007 mV ($p = 0.38$). This trend suggests that Rectus Femoris activation may decrease slightly with increasing speed, in the presence of the knee brace.

At all speeds, the Tibialis Anterior consistently showed decreased activation with the brace. The RMS decreased by -0.038 mV at 2Km/H, -0.023 mV at 4Km/H, and -0.018 mV at 6Km/H. The reductions are consistent, but the magnitude of change decreases as walking speed increases.

The knee brace increased activation of the Vastus Lateralis at all speeds. At 2 km/H, the RMS difference was 0.025 mV ($p < 0.05$), and this increased slightly to 0.030 mV at 4 km/H ($p < 0.05$). The increase was smaller but still positive (0.018 mV) ($p = 0.17$) at 6Km/H. Results of these experiments suggest that the Vastus Lateralis serves a role when walking with a knee brace.

Finally, the Vastus Medialis demonstrated increased activation with the brace at slower speeds and little difference at faster speeds. The RMS increased by 0.079 mV at 2Km/H ($p < 0.05$) and by 0.046 mV at 4Km/H ($p < 0.05$). At 6Km/H though, the difference was almost negligible (-0.006 mV) ($p = 0.67$). Thus, the knee brace has the greatest effect on Vastus Medialis activation at lower speeds.

Table 1
Overall Summary For The RMS Results Of All Muscles, Across All Speeds. All Values Are Reported In mV.

Muscle	2Km/H			4Km/H			6Km/H		
	W	W/O	Diff	W	W/O	Diff	W	W/O	Diff
Biceps Femoris	0.351	0.277	0.074	0.355	0.325	0.030	0.396	0.369	0.026

Table 1 (*cont'd*)

Gastrocnemius Lateralis	0.261	0.341	-0.080	0.372	0.370	0.002	0.386	0.433	-0.047
Gastrocnemius Medialis	0.295	0.300	-0.005	0.388	0.387	0.001	0.400	0.420	-0.020
Rectus Femoris	0.289	0.298	-0.009	0.370	0.397	-0.027	0.366	0.373	-0.007
Tibialis Anterior	0.355	0.393	-0.038	0.418	0.441	-0.023	0.445	0.463	-0.018
Vastus Lateralis	0.348	0.323	0.025	0.383	0.353	0.030	0.427	0.408	0.018
Vastus Medialis	0.293	0.214	0.079	0.367	0.321	0.046	0.428	0.434	-0.006

4.1.2 Peak muscle amplitude. Biceps Femoris Across all walking speeds, the knee brace increased peak activation of the Biceps Femoris muscle. The peak EMG value increased from 0.479 mV without the brace to 0.547 mV with the brace, an increase of 0.068 mV at 2Km/H. The difference was smaller at 4Km/H, 0.001 mV (0.551 mV with the brace and 0.550 mV without). The difference was minimal at 6Km/H (0.001 mV) with peak values of 0.592 mV (with brace) and 0.591 mV (without). These results show that the brace slightly increases peak activation of the Biceps Femoris, but the effect diminishes with increasing walking speed.

Gastrocnemius Lateralis Peak activation of the Gastrocnemius Lateralis was reduced with the brace, especially at slower speeds. The RMS peak value decreased from 0.552 mV without the brace to 0.415 mV with the brace, a reduction of -0.137 mV at 2 km/H. At 4Km/H the decrease was smaller, with a difference of -0.012 mV (0.579 mV with brace vs. 0.579 mV without). The difference at 6Km/H was -0.082 mV (0.562 mV with brace vs. 0.644 mV without). These findings indicate that the brace decreases Gastrocnemius Lateralis activation, most at slower speeds, with a decreasing effect as speed increases.

Gastrocnemius Medialis Consistent reductions in the peak activation of the Gastrocnemius Medialis were observed with the brace at all speeds. The difference was -0.031 mV (0.508 mV with the brace, 0.540 mV without) at 2 km/H. The reduction

was -0.033 mV (0.635 mV with brace vs. 0.602 mV without) at 4 km/H. The decrease was -0.047 mV (0.582 mV with brace vs. 0.629 mV without) at 6 km/H. These results indicate a small decrease in Gastrocnemius Medialis activation at all walking speeds.

Rectus Femoris The results for the Rectus Femoris were mixed. The peak activation decreased slightly to 0.405 mV (-0.005 mV) with the brace at 2Km/H, compared to 0.410 mV without the brace. The brace reduced activation by -0.053 mV at 4Km/H (0.548 mV with the brace and 0.601 mV without). At 6Km/H, the difference was -0.040 mV (0.489 mV with the brace and 0.529 mV without). These results indicate that the brace decreases Rectus Femoris activation at higher speeds, but has no effect at slower speeds.

Tibialis Anterior The Tibialis Anterior showed reduced peak activation with the brace. The peak value decreased by -0.075 mV (0.643 mV without the brace; 0.568 mV with the brace) at 2 km/H. The reduction was -0.080 mV at 4Km/H (0.685 mV without the brace, 0.605 mV with the brace). The decrease was -0.064 mV (0.692 mV without the brace vs. 0.628 mV with the brace) at 6 km/H. These results show that the knee brace reduces Tibialis Anterior activation at all speeds, but the magnitude of reduction decreases with increasing walking speed

Vastus Lateralis The peak activation of the Vastus Lateralis was consistent with increases with the brace. The difference at 2Km/H was 0.083 mV (0.567 mV with the brace vs. 0.484 mV without). The increase was 0.094 mV (0.607 mV with brace vs. 0.513 mV without) at 4Km/H. The increase was slightly lower at 6Km/H, 0.016 mV (0.608 mV with brace vs. 0.592 mV without). The results indicate that the brace consistently increases Vastus Lateralis activation, but the magnitude of change is small.

Vastus Medialis With the brace, the Vastus Medialis also showed increased peak activation, especially at slower speeds. The RMS peak increased from 0.312 mV without the brace to 0.494 mV with the brace, a difference of 0.079 mV at 2 km/H. The increase was smaller at 0.064 mV (0.582 mV with brace vs. 0.518 mV without) at 4 km/H. The change at 6Km/H was -0.055 mV (0.603 mV with brace vs. 0.658 mV

without). These results indicate that the brace has a greater effect on Vastus Medialis activation at slower speeds.

Table 2

Overall Summary For The Median Of The Peak Values Of All Muscles' Activities, Across All Speeds. All Values Are Reported In mV.

Muscle	2Km/H			4Km/H			6Km/H		
	W	W/O	Diff	W	W/O	Diff	W	W/O	Diff
Biceps Femoris	0.547	0.479	0.068	0.551	0.550	0.001	0.592	0.591	0.001
Gastrocnemius Lateralis	0.415	0.552	-0.137	0.567	0.579	-0.012	0.562	0.644	-0.082
Gastrocnemius Medialis	0.508	0.540	-0.031	0.602	0.635	-0.033	0.582	0.629	-0.047
Rectus Femoris	0.405	0.410	-0.005	0.548	0.601	-0.053	0.489	0.529	-0.040
Tibialis Anterior	0.568	0.643	-0.075	0.605	0.685	-0.080	0.628	0.692	-0.064
Vastus Lateralis	0.567	0.484	0.083	0.607	0.513	0.094	0.608	0.592	0.016
Vastus Medialis	0.494	0.312	0.079	0.582	0.518	0.064	0.603	0.658	-0.055

4.1.3 Peak Amplitude at initial contact. The Biceps Femoris demonstrated increased peak activation with the knee brace at all walking speeds during initial contact. At 2Km/H, the peak increased from 0.357 mV without the brace to 0.417 mV with the brace, a difference of 0.060 mV ($p < 0.05$). At 4Km/H, the peak decreased slightly but still showed a small increase of 0.018 mV (W: 0.472 mV, W/O: 0.485 mV ($p = 0.75$)). At 6 km/H, the difference was minimal (0.005 mV) ($p = 0.13$). These results indicate that the brace has a greater impact on Biceps Femoris activation at slower speeds, with diminishing effects as speed increases.

The Gastrocnemius Lateralis showed mixed responses to the brace. At 2Km/H, the peak decreased from 0.196 mV to 0.171 mV, resulting in a difference of -0.025 mV ($p < 0.05$). However, at 4Km/H, the peak increased with the brace (W: 0.264 mV, W/O: 0.228 mV), a positive difference of 0.036 mV ($p < 0.05$). At 6Km/H, there was again a slight increase of 0.013 mV ($p < 0.05$). These results suggest that the brace

reduces activation at slower speeds but may facilitate slightly higher activation at moderate and faster speeds.

The Gastrocnemius Medialis showed consistently higher activation with the brace across all speeds. At 2Km/H, the peak value increased from 0.095 mV without the brace to 0.151 mV with the brace, a difference of 0.055 mV ($p < 0.05$). At 4Km/H, the difference grew to 0.068 mV (W: 0.255 mV, W/O: 0.187 mV) ($p < 0.05$), and at 6Km/H, the brace increased the peak to 0.340 mV compared to 0.284 mV without the brace (0.056 mV difference) ($p < 0.05$). This consistent increase suggests that the brace leads to higher Gastrocnemius Medialis activation during initial contact.

In contrast, the Rectus Femoris showed a reduction in peak activation with the brace at higher walking speeds. At 2Km/H, the difference was small (-0.018 mV) ($p < 0.05$), with values of 0.308 mV (W) and 0.326 mV (W/O). At 4Km/H, the reduction was more pronounced (-0.042 mV) ($p < 0.05$), with a peak of 0.455 mV with the brace and 0.498 mV without it. At 6 km/H, the values were nearly identical, with a difference of -0.002 mV ($p < 0.05$). These results indicate that the brace slightly reduces Rectus Femoris activation, particularly at moderate speeds.

The Tibialis Anterior consistently showed reduced activation with the brace across all speeds. At 2Km/H, the peak decreased significantly from 0.370 mV to 0.266 mV, a difference of -0.104 mV ($p < 0.05$). At 4Km/H, the reduction continued (W: 0.355 mV, W/O: 0.453 mV, Diff: -0.098 mV) ($p < 0.05$), and at 6Km/H, the brace again reduced the peak from 0.471 mV to 0.374 mV, a difference of -0.097 mV ($p < 0.05$). These results suggest that the brace consistently reduces Tibialis Anterior activation during initial contact, regardless of walking speed.

The Vastus Lateralis displayed consistently higher activation with the brace across all speeds. At 2Km/H, the peak increased from 0.369 mV to 0.438 mV, a difference of 0.069 mV ($p < 0.05$). At 4Km/H, the increase was slightly larger (0.076 mV, W: 0.504 mV, W/O: 0.428 mV) ($p < 0.05$), and at 6Km/H, the difference was 0.015 mV ($p = 0.19$). These results suggest that the brace enhances Vastus Lateralis activation, with the greatest effect at slower to moderate speeds.

The Vastus Medialis showed mixed effects. At 2Km/H, the brace increased the peak significantly, from 0.246 mV to 0.375 mV, a difference of 0.128 mV ($p < 0.05$). At 4Km/H, the increase was smaller (0.058 mV, W: 0.485 mV, W/O: 0.428 mV) ($p <$

0.05). However, at 6Km/H, the peak activation decreased slightly with the brace, from 0.579 mV to 0.516 mV (-0.062 mV difference) ($p < 0.05$). These findings suggest that the brace increases Vastus Medialis activation at slower speeds but reduces it slightly at faster speeds.

Table 3

Overall Summary For The Median Of The Peak Values Of All Muscles' Activities At Initial Contact, Across All Speeds. All Values Are Reported In mV.

Muscle	2Km/H			4Km/H			6Km/H		
	W	W/O	Diff	W	W/O	Diff	W	W/O	Diff
Biceps Femoris	0.417	0.357	0.060	0.472	0.485	0.018	0.494	0.488	0.005
Gastrocnemius Lateralis	0.171	0.196	-0.025	0.264	0.228	0.036	0.275	0.262	0.013
Gastrocnemius Medialis	0.151	0.095	0.055	0.255	0.187	0.068	0.340	0.284	0.056
Rectus Femoris	0.308	0.326	-0.018	0.455	0.498	-0.042	0.421	0.423	-0.002
Tibialis Anterior	0.266	0.370	-0.104	0.355	0.453	-0.098	0.374	0.471	-0.097
Vastus Lateralis	0.438	0.369	0.069	0.504	0.428	0.076	0.530	0.515	0.015
Vastus Medialis	0.375	0.246	0.128	0.485	0.428	0.058	0.516	0.579	-0.062

4.1.4 Peak Amplitude during Loading Response. The Biceps Femoris showed consistently higher peak activation with the brace during the loading response phase. At 2Km/H, the peak value increased from 0.367 mV without the brace to 0.442 mV with the brace, resulting in a difference of 0.075 mV ($p < 0.05$). At 4Km/H, the increase was smaller, with a difference of 0.024 mV (W: 0.480 mV, W/O: 0.456 mV) ($p = 0.22$). At 6Km/H, the difference was minimal, with values of 0.496 mV and 0.490 mV (Diff: 0.006 mV) ($p = 0.30$). These results suggest that the brace increases Biceps Femoris activation, particularly at slower walking speeds, but this effect diminishes as speed increases.

The Gastrocnemius Lateralis exhibited reduced peak activation with the brace at slower speeds. At 2Km/H, the peak value decreased significantly from 0.282 mV to 0.212 mV, a difference of -0.069 mV ($p < 0.05$). However, at 4Km/H, the brace led to a slight increase (W: 0.340 mV, W/O: 0.307 mV, Diff: 0.032 mV) ($p < 0.05$). At 6 km/H, the difference was negligible (0.003 mV) ($p = 0.054$). These findings indicate that the brace reduces Gastrocnemius Lateralis activation at slower speeds but has minimal or slight positive effects at higher speeds.

The Gastrocnemius Medialis demonstrated increased peak activation with the brace across all speeds. At 2Km/H, the brace resulted in a significant increase from 0.148 mV to 0.209 mV (Diff: 0.061 mV) ($p < 0.05$). At 4Km/H, the increase was smaller, with a difference of 0.001 mV (W: 0.353 mV, W/O: 0.292 mV) ($p < 0.05$). At 6Km/H, the difference was 0.013 mV ($p < 0.05$), with values of 0.446 mV (W) and 0.433 mV (W/O). These results suggest that the brace consistently increases Gastrocnemius Medialis activation, particularly at slower speeds.

The Rectus Femoris showed reduced peak activation with the brace at moderate speeds. At 2 km/H, the difference was minimal (-0.010 mV) ($p = 0.20$), with values of 0.333 mV (W) and 0.343 mV (W/O). At 4Km/H, the reduction was larger, with the peak decreasing from 0.517 mV to 0.480 mV (Diff: -0.027 mV) ($p < 0.05$). At 6 km/H, the difference was small (0.003 mV) ($p = 0.11$). These findings suggest that the brace has a minimal effect at slower and faster speeds but reduces Rectus Femoris activation at moderate speeds.

The Tibialis Anterior consistently exhibited reduced activation with the brace across all speeds. At 2Km/H, the peak value decreased significantly from 0.378 mV to 0.282 mV, a difference of -0.096 mV ($p < 0.05$). At 4Km/H, the brace reduced activation from 0.456 mV to 0.359 mV (Diff: -0.023 mV) ($p < 0.05$). At 6Km/H, the reduction persisted, with values of 0.476 mV (W/O) and 0.379 mV (W), resulting in a difference of -0.097 mV ($p < 0.05$). These results indicate that the brace consistently reduces Tibialis Anterior activation during the loading response phase.

The Vastus Lateralis showed increased activation with the brace across all speeds. At 2Km/H, the peak increased from 0.390 mV without the brace to 0.472 mV with the brace (Diff: 0.082 mV) ($p < 0.05$). At 4Km/H, the increase was smaller but

still positive, with values of 0.515 mV (W) and 0.434 mV (W/O), resulting in a difference of 0.030 mV ($p < 0.05$). At 6Km/H, the brace led to a slight increase of 0.023 mV (W: 0.543 mV, W/O: 0.521 mV) ($p = 0.84$). These results suggest that the brace consistently enhances Vastus Lateralis activation during loading response.

Finally, the Vastus Medialis demonstrated mixed responses to the brace. At 2Km/H, the peak activation increased significantly from 0.253 mV without the brace to 0.392 mV with the brace, a difference of 0.139 mV ($p < 0.05$). At 4Km/H, the increase was smaller (W: 0.490 mV, W/O: 0.433 mV, Diff: 0.046 mV) ($p < 0.05$). However, at 6Km/H, the brace resulted in a reduction, with the peak decreasing from 0.581 mV to 0.530 mV (Diff: -0.051 mV) ($p < 0.05$). These findings suggest that the brace increases Vastus Medialis activation at slower and moderate speeds but reduces it slightly at faster speeds.

Table 4

Overall Summary For The Median Of The Peak Values Of All Muscles' Activities During The Loading Response Phase, Across All Speeds. All Values Are Reported In mV.

Muscle	2Km/H			4Km/H			6Km/H		
	W	W/O	Diff	W	W/O	Diff	W	W/O	Diff
Biceps Femoris	0.442	0.367	0.075	0.480	0.456	0.024	0.496	0.490	0.006
Gastrocnemius Lateralis	0.212	0.282	-0.069	0.340	0.307	0.032	0.389	0.386	0.003
Gastrocnemius Medialis	0.209	0.148	0.061	0.353	0.292	0.001	0.446	0.433	0.013
Rectus Femoris	0.333	0.343	-0.010	0.480	0.517	-0.027	0.435	0.432	0.003
Tibialis Anterior	0.282	0.378	-0.096	0.359	0.456	-0.023	0.379	0.476	-0.097
Vastus Lateralis	0.472	0.390	0.082	0.515	0.434	0.030	0.543	0.521	0.023
Vastus Medialis	0.392	0.253	0.139	0.490	0.433	0.046	0.530	0.581	-0.051

4.1.5 Variability in EMG amplitude over the gait cycle (CoV). The Biceps Femoris showed consistently lower variability in EMG amplitude with the brace across all walking speeds. At 2Km/H, the CoV value decreased from 0.407 without the brace to 0.308 with the brace, a difference of -0.099 ($p < 0.05$). At 4Km/H, the variability reduction was larger, with a difference of -0.178 (W: 0.344, W/O: 0.522) ($p < 0.05$). At 6Km/H, the reduction was -0.125 (W: 0.372, W/O: 0.497) ($p < 0.05$). These results suggest that the knee brace reduces variability in Biceps Femoris activation, with the largest effect observed at moderate walking speeds.

The Gastrocnemius Lateralis also exhibited reduced variability with the brace across all speeds. At 2Km/H, the CoV value decreased significantly by -0.153 (W: 0.410, W/O: 0.563) ($p < 0.05$). At 4Km/H, the reduction was smaller (-0.077), with values of 0.427 (W) and 0.504 (W/O) ($p < 0.05$). At 6 km/H, the variability decreased by -0.109 (W: 0.393, W/O: 0.502) ($p < 0.05$). These findings indicate that the brace reduces variability in Gastrocnemius Lateralis activation, with the most pronounced effect at slower speeds.

For the Gastrocnemius Medialis, the CoV values were consistently lower with the brace. At 2Km/H, the variability decreased significantly from 0.714 without the brace to 0.495 with the brace, a difference of -0.219 ($p < 0.05$). At 4Km/H, the reduction was -0.139 (W: 0.448, W/O: 0.587) ($p < 0.05$), and at 6Km/H, the CoV decreased by -0.168 (W: 0.340, W/O: 0.508) ($p < 0.05$). These results suggest that the brace has a substantial stabilizing effect on Gastrocnemius Medialis variability, especially at slower speeds.

The Rectus Femoris displayed a small but consistent reduction in variability with the brace. At 2Km/H, the CoV increased slightly with the brace (0.098 difference), with values of 0.263 (W) and 0.165 (W/O) ($p < 0.05$). At 4Km/H, the variability reduction was minimal (-0.027) ($p < 0.05$), while at 6Km/H, the brace reduced the CoV from 0.279 to 0.201, a difference of -0.078 ($p < 0.05$). These results suggest that the brace has a relatively small effect on Rectus Femoris variability, particularly at higher walking speeds.

The Tibialis Anterior also demonstrated reduced variability with the brace across all speeds. At 2Km/H, the CoV decreased from 0.496 to 0.435, a difference of -0.060 ($p < 0.05$). At 4Km/H, the reduction was slightly larger (-0.064, W: 0.354, W/O: 0.418) ($p < 0.05$). At 6Km/H, the difference was -0.049 with values of 0.276 (W) and 0.325 (W/O) ($p < 0.05$). These findings indicate that the knee brace consistently reduces variability in Tibialis Anterior activation across all speeds, though the magnitude of change is relatively small.

Table 5

Overall Summary For The Median Of The Cov Values Of All Muscles' Activities During The Loading Response Phase, Across All Speeds.

Muscle	2Km/H			4Km/H			6Km/H		
	W	W/O	Diff	W	W/O	Diff	W	W/O	Diff
Biceps Femoris	0.308	0.407	-0.099	0.344	0.522	-0.178	0.372	0.497	-0.125
Gastrocnemius Lateralis	0.410	0.563	-0.153	0.427	0.504	-0.077	0.393	0.502	-0.109
Gastrocnemius Medialis	0.495	0.714	-0.219	0.448	0.587	-0.139	0.340	0.508	-0.168
Rectus Femoris	0.263	0.165	0.098	0.249	0.276	-0.027	0.201	0.279	-0.078
Tibialis Anterior	0.435	0.496	-0.060	0.354	0.418	-0.064	0.276	0.325	-0.049
Vastus Lateralis	0.367	0.214	0.153	0.364	0.364	0.000	0.307	0.480	-0.172
Vastus Medialis	0.312	0.228	0.084	0.318	0.352	-0.034	0.311	0.449	-0.138

The Vastus Lateralis displayed mixed results, with variability initially increasing slightly at slower speeds but decreasing at faster speeds. At 2Km/H, the CoV increased from 0.214 without the brace to 0.367 with the brace, a difference of 0.153 ($p < 0.05$). At 4Km/H, no difference was observed between the conditions (W: 0.364, W/O: 0.364) ($p < 0.05$). At 6Km/H, the variability decreased by -0.172, with values of 0.30

7 (W) and 0.480 (W/O) ($p < 0.05$). These results suggest that the brace reduces variability at faster speeds but may introduce greater variability at slower speeds.

The Vastus Medialis exhibited reduced variability with the brace across all speeds. At 2Km/H, the CoV increased slightly (0.084 difference, W: 0.312, W/O: 0.228) ($p < 0.05$). At 4Km/H, the reduction was minimal (-0.034, W: 0.318, W/O: 0.352) ($p < 0.05$). At 6 km/H, the variability decreased by -0.138, with values of 0.311 (W) and 0.449 (W/O) ($p < 0.05$). These results suggest that the brace reduces variability in Vastus Medialis activation, particularly at higher walking speeds.

4.1.6 Average of the absolute values of the EMG signal (MAV). The Biceps Femoris showed consistently higher MAV values with the knee brace across all walking speeds. At 2Km/H, the MAV increased from 0.252 mV without the brace to 0.329 mV with the brace, a difference of 0.076 mV ($p < 0.05$). At 4Km/H, the increase was smaller, with values of 0.321 mV (W) and 0.284 mV (W/O), resulting in a difference of 0.037 mV ($p < 0.05$). At 6Km/H, the MAV increased slightly, with a difference of 0.040 mV (W: 0.363 mV, W/O: 0.323 mV) ($p < 0.05$). These results suggest that the brace consistently increases Biceps Femoris activation during the loading response phase, particularly at slower speeds.

The Gastrocnemius Lateralis demonstrated reduced MAV values with the brace at slower and faster speeds. At 2Km/H, the MAV decreased from 0.290 mV to 0.231 mV, a difference of -0.059 mV ($p < 0.05$). At 4Km/H, the difference was negligible (0.004 mV), with values of 0.331 mV (W) and 0.327 mV (W/O) ($p = 0.22$). At 6 km/H, the brace again led to a reduction in MAV (-0.034 mV), with values of 0.353 mV (W) and 0.387 mV (W/O) ($p < 0.05$). These findings suggest that the brace reduces Gastrocnemius Lateralis activation, particularly at slower and higher speeds.

For the Gastrocnemius Medialis, the MAV values remained largely unaffected by the brace. At 2Km/H, the difference was minimal (0.003 mV) ($p < 0.05$), with values of 0.243 mV (W) and 0.240 mV (W/O). At 4Km/H, the MAV increased slightly with the brace (0.014 mV) ($p < 0.05$), but no difference was observed at 6Km/H, where

both conditions resulted in an MAV of 0.371 mV ($p < 0.05$). These results indicate that the brace has little to no effect on Gastrocnemius Medialis activation.

The Rectus Femoris exhibited minimal changes in MAV values with the brace. At 2Km/H, the MAV increased slightly from 0.266 mV to 0.271 mV, a difference of 0.005 mV ($p = 0.43$). At 4Km/H, the MAV decreased slightly (-0.015 mV) ($p < 0.05$), with values of 0.345 mV (W) and 0.360 mV (W/O). At 6Km/H, the reduction was negligible (-0.004 mV) ($p = 0.33$), with values of 0.348 mV (W) and 0.352 mV (W/O). These results suggest that the brace has minimal influence on Rectus Femoris activation.

The Tibialis Anterior displayed slightly reduced MAV values with the brace across all speeds. At 2Km/H, the MAV decreased from 0.340 mV without the brace to 0.321 mV with the brace (-0.019 mV) ($p < 0.05$). At 4Km/H, the reduction was slightly larger (-0.011 mV) ($p < 0.05$), with values of 0.389 mV (W) and 0.400 mV (W/O). At 6Km/H, the MAV decreased marginally by -0.007 mV ($p < 0.05$), with values of 0.426 mV (W) and 0.433 mV (W/O). These results suggest that the brace slightly reduces Tibialis Anterior activation.

Table 6
Overall Summary For The Median Of The Mav Values Of All Muscles' Activities During The Loading Response Phase, Across All Speeds. All Values Are Reported In mV.

Muscle	2Km/H			4Km/H			6Km/H		
	W	W/O	Diff	W	W/O	Diff	W	W/O	Diff
Biceps Femoris	0.329	0.252	0.076	0.321	0.284	0.037	0.363	0.323	0.040
Gastrocnemius Lateralis	0.231	0.290	-0.059	0.331	0.327	0.004	0.353	0.387	-0.034
Gastrocnemius Medialis	0.243	0.240	0.003	0.343	0.329	0.014	0.371	0.371	0.000
Rectus Femoris	0.271	0.266	0.005	0.345	0.360	-0.015	0.348	0.352	-0.004

Table 6 (cont'd)

Tibialis Anterior	0.321	0.340	-0.019	0.389	0.400	-0.011	0.426	0.433	-0.007
Vastus Lateralis	0.312	0.289	0.023	0.345	0.313	0.032	0.398	0.366	0.032
Vastus Medialis	0.266	0.202	0.064	0.329	0.278	0.051	0.403	0.384	0.019

The Vastus Lateralis showed consistent increases in MAV values with the brace across all speeds. At 2Km/H, the MAV increased from 0.289 mV to 0.312 mV, a difference of 0.023 mV ($p = 0.06$). At 4Km/H, the increase was 0.032 mV (W: 0.345 mV, W/O: 0.313 mV) ($p < 0.05$). At 6Km/H, the MAV increased further, with a difference of 0.032 mV (W: 0.398 mV, W/O: 0.366 mV) ($p < 0.05$). These findings suggest that the brace enhances Vastus Lateralis activation across all walking speeds.

The Vastus Medialis also demonstrated higher MAV values with the brace across all speeds. At 2Km/H, the MAV increased from 0.202 mV without the brace to 0.266 mV with the brace, a difference of 0.064 mV ($p < 0.05$). At 4Km/H, the difference was 0.051 mV, with values of 0.329 mV (W) and 0.278 mV (W/O) ($p < 0.05$). At 6Km/H, the MAV increased slightly (0.019 mV), with values of 0.403 mV (W) and 0.384 mV (W/O) ($p < 0.05$). These results suggest that the brace consistently increases Vastus Medialis activation.

4.2 Lower Body Kinematics Results

In this study, we analyzed key kinematic parameters, Root Mean Square (RMS), Range of Motion (ROM), and maximum and minimum joint angles, to assess the effect of wearing a KAFO brace during gait at different speeds (2KMH, 4KMH, 6KMH). The parameters were chosen to capture both dynamic and static aspects of joint behavior and to give insight into the biomechanical effects of bracing on gait.

These parameters as a whole give a complete picture of the effect of the knee brace on movement kinematics, joint stability, and gait economy. All results are presented in median form as the data is not normally distributed, in order to provide a more stable measure of central tendency. For completeness, additional analyses of

other kinematic metrics can be found in the Appendix section. The findings regarding the kinematic parameters are summarised in the tables below.

4.2.1 RMS results for joint angles. The Right Ankle Abduction/Adduction exhibited minimal changes in RMS values with the brace at slower and moderate speeds, but a significant increase at higher speeds. At 2Km/H, the RMS value decreased slightly from 6.311° without the brace to 6.248° with the brace, a difference of -0.063° ($p < 0.05$). At 4Km/H, the brace led to a slight increase of 0.083° (W: 7.157° , W/O: 7.074°) ($p = 0.10$). At 6Km/H, the RMS value increased significantly by 1.344° (W: 8.685° , W/O: 7.341°) ($p < 0.05$). These results indicate that the brace has a greater effect on ankle abduction/adduction at higher speeds.

The Right Hip Abduction/Adduction showed decreased RMS values with the brace across all speeds. At 2Km/H, the RMS decreased by -0.784° (W: 3.351° , W/O: 4.135°) ($p < 0.05$). At 4Km/H, the reduction was more pronounced (-1.348°), with values of 3.916° (W) and 5.264° (W/O) ($p < 0.05$). At 6Km/H, the reduction was -0.857° (W: 4.853° , W/O: 5.710°) ($p < 0.05$). These results suggest that the brace reduces hip abduction/adduction movement across all speeds, particularly at moderate speeds.

The Right Knee Abduction/Adduction showed mixed results. At 2Km/H, the RMS value increased slightly with the brace by 0.324° (W: 2.089° , W/O: 1.764°) ($p < 0.05$). However, at 4Km/H, the brace caused a reduction of -0.347° (W: 1.830° , W/O: 2.176°) ($p < 0.05$). At 6Km/H, the reduction was -0.575° (W: 1.928° , W/O: 2.502°) ($p < 0.05$). These findings indicate that the brace initially increases knee abduction/adduction at slower speeds but restricts it at higher speeds.

The Right Ankle Flexion/Extension demonstrated consistently reduced RMS values with the brace across all speeds. At 2Km/H, the RMS decreased from 6.962° without the brace to 5.835° with the brace, a difference of -1.127° ($p < 0.05$). At 4Km/H, the reduction increased to -2.366° (W: 6.290° , W/O: 8.657°) ($p < 0.05$). At 6Km/H, the brace caused the largest reduction of -3.072° (W: 7.079° , W/O: 10.152°)

($p < 0.05$). These results suggest that the brace significantly limits ankle flexion/extension, particularly at faster speeds.

The Right Hip Flexion/Extension showed higher RMS values with the brace across all speeds. At 2Km/H, the RMS increased by 1.180° (W: 12.892° , W/O: 11.712°) ($p < 0.05$). At 4Km/H, the increase was smaller (1.210° , W: 14.950° , W/O: 13.740°) ($p < 0.05$), and at 6Km/H, the brace caused an increase of 0.778° (W: 16.375° , W/O: 15.597°) ($p < 0.05$). These findings suggest that the brace facilitates greater hip flexion/extension movements during gait.

The Right Knee Flexion/Extension exhibited consistently higher RMS values with the brace. At 2Km/H, the RMS increased by 3.050° (W: 25.969° , W/O: 22.919°) ($p < 0.05$). At 4Km/H, the difference was smaller (1.047°), with values of 28.799° (W) and 27.752° (W/O) ($p < 0.05$). At 6Km/H, the increase was 0.678° (W: 29.621° , W/O: 28.943°) ($p < 0.05$). These results indicate that the brace enhances knee flexion/extension, particularly at slower speeds.

The Right Ankle Internal/External Rotation showed reduced RMS values with the brace across all speeds. At 2Km/H, the RMS decreased from 6.953° without the brace to 3.694° with the brace, a difference of -3.258° ($p < 0.05$). At 4Km/H, the reduction was -3.351° (W: 4.019° , W/O: 7.370°) ($p = 5.8$). At 6Km/H, the brace caused a decrease of -1.943° (W: 5.411° , W/O: 7.354°) ($p < 0.05$). These findings suggest that the brace significantly restricts ankle internal/external rotation, particularly at slower speeds.

Table 7

Summary Of The Rms Of The Angles' Values Of All Joints, Across All Speeds. All Values Are Reported In Degrees.

Joints	2Km/H			4Km/H			6Km/H		
	W	W/O	Diff	W	W/O	Diff	W	W/O	Diff
jRight Ankle Abd/Add	6.248	6.311	-0.063	7.157	7.074	0.083	8.685	7.341	1.344

Table 7 (cont'd)

jRight Hip Abd/Add	3.351	4.135	-0.784	3.916	5.264	-1.348	4.853	5.710	-0.857
jRight Knee Abd/Add	2.089	1.764	0.324	1.830	2.176	-0.347	1.928	2.502	-0.575
jRight Ankle Flex/Ext	5.835	6.962	-1.127	6.290	8.657	-2.366	7.079	10.152	-3.072
jRight Hip Flex/Ext	12.892	11.712	1.180	14.950	13.740	1.210	16.375	15.597	0.778
jRight Knee Flex/Ext	25.969	22.919	3.050	28.799	27.752	1.047	29.621	28.943	0.678
jRight Ankle Int/Ext	3.694	6.953	-3.258	4.019	7.370	-3.351	5.411	7.354	-1.943
jRight Hip Int/Ext	4.889	3.974	0.915	4.751	4.211	0.540	4.759	4.375	0.384
jRight Knee Int/Ext	3.077	2.896	0.181	3.005	3.485	-0.480	2.820	3.996	-1.175

The Right Hip Internal/External Rotation demonstrated increased RMS values with the brace across all speeds. At 2Km/H, the RMS increased by 0.915° (W: 4.889°, W/O: 3.974°) ($p < 0.05$). At 4Km/H, the increase was smaller (0.540°), with values of 4.751° (W) and 4.211° (W/O) ($p < 0.05$). At 6Km/H, the difference was 0.384° (W: 4.759°, W/O: 4.375°) ($p < 0.05$). These findings indicate that the brace slightly increases hip internal/external rotation.

Finally, the Right Knee Internal/External Rotation showed mixed effects. At 2Km/H, the RMS increased slightly with the brace (0.181°, W: 3.077°, W/O: 2.896°) ($p < 0.05$). At 4Km/H, the brace led to a reduction of -0.480° (W: 3.005°, W/O: 3.485°) ($p < 0.05$). At 6Km/H, the RMS decreased further by -1.175° (W: 2.820°, W/O: 3.996°) ($p < 0.05$). These results suggest that the brace reduces knee internal/external rotation at higher speeds.

4.2.2 ROM results for joint angles. The Right Ankle Abduction/Adduction demonstrated a substantial reduction in ROM with the brace across all walking speeds. At 2Km/H, the ROM decreased from 13.983° without the brace to 7.858° with the brace, a difference of -6.125° ($p < 0.05$). At 4Km/H, the reduction was similar (-6.538°, W: 7.996°, W/O: 14.533°) ($p < 0.05$). At 6Km/H, the ROM decreased further to -

7.119° (W: 8.957°, W/O: 16.076°) ($p < 0.05$). These results indicate that the knee brace significantly restricts ankle abduction/adduction, with the restriction increasing at faster speeds.

The Right Hip Abduction/Adduction also exhibited reduced ROM with the brace at all speeds. At 2Km/H, the ROM decreased from 12.203° without the brace to 8.579° with the brace, a difference of -3.623° ($p < 0.05$). At 4Km/H, the ROM was further reduced by -3.686° (W: 10.919°, W/O: 14.605°) ($p < 0.05$). At 6Km/H, the reduction was -3.027° (W: 13.237°, W/O: 16.263°) ($p < 0.05$). These results suggest that the brace limits hip abduction/adduction motion, although the restriction decreases slightly as speed increases.

The Right Knee Abduction/Adduction showed consistent reductions in ROM with the brace. At 2Km/H, the ROM decreased by -0.761° (W: 5.843°, W/O: 6.604°) ($p < 0.05$). At 4Km/H, the reduction increased to -3.375° (W: 5.183°, W/O: 8.558°) ($p < 0.05$). At 6Km/H, the ROM decreased further by -3.853° (W: 6.087°, W/O: 9.941°) ($p < 0.05$). These results indicate that the brace significantly restricts knee abduction/adduction, particularly at faster speeds.

The Right Ankle Flexion/Extension showed the largest reductions in ROM with the brace across all speeds. At 2Km/H, the ROM decreased from 21.658° without the brace to 12.400° with the brace, a difference of -9.258° ($p < 0.05$). At 4Km/H, the reduction increased to -13.211° (W: 18.129°, W/O: 31.340°) ($p < 0.05$). At 6Km/H, the ROM decreased further by -13.006° (W: 23.307°, W/O: 36.312°) ($p < 0.05$). These findings highlight that the brace substantially limits ankle flexion/extension motion, particularly at higher speeds.

The Right Hip Flexion/Extension showed a slight increase in ROM with the brace at all speeds. At 2Km/H, the ROM increased by 1.500° (W: 31.286°, W/O: 29.786°) ($p < 0.05$). At 4Km/H, the ROM increased by 1.758° (W: 38.468°, W/O: 36.710°) ($p < 0.05$). At 6Km/H, the increase was 1.400° (W: 44.012°, W/O: 42.612°)

($p < 0.05$). These results suggest that the brace slightly enhances hip flexion/extension motion, particularly at slower to moderate speeds.

The Right Knee Flexion/Extension exhibited consistent reductions in ROM with the brace. At 2Km/H, the ROM decreased by -2.234° (W: 51.115° , W/O: 53.349°) ($p < 0.05$). At 4Km/H, the reduction was larger (-4.916° , W: 56.866° , W/O: 61.782°) ($p < 0.05$). At 6Km/H, the ROM decreased further by -5.523° (W: 57.297° , W/O: 62.819°) ($p < 0.05$). These findings indicate that the knee brace restricts knee flexion/extension motion across all speeds, with the greatest restriction observed at higher speeds.

The Right Ankle Internal/External Rotation showed significant reductions in ROM with the brace. At 2Km/H, the ROM decreased from 15.440° to 8.961° , a difference of -6.479° ($p < 0.05$). At 4Km/H, the reduction was -9.015° (W: 10.304° , W/O: 19.319°) ($p < 0.05$). At 6Km/H, the ROM decreased by -7.340° (W: 15.186° , W/O: 22.527°) ($p < 0.05$). These results suggest that the brace substantially limits ankle internal/external rotation, particularly at moderate speeds.

The Right Hip Internal/External Rotation demonstrated slight increases in ROM with the brace. At 2Km/H, the ROM increased by 3.385° (W: 12.701° , W/O: 9.317°) ($p < 0.05$). At 4Km/H, the increase was smaller (2.380° , W: 12.008° , W/O: 9.627°) ($p < 0.05$). At 6Km/H, the ROM increased by 1.438° (W: 11.989° , W/O: 10.550°) ($p < 0.05$). These results suggest that the brace enhances hip internal/external rotation slightly at slower to moderate speeds.

The Right Knee Internal/External Rotation showed reductions in ROM with the brace across all speeds. At 2Km/H, the ROM decreased by -1.020° (W: 8.790° , W/O: 9.810°) ($p < 0.05$). At 4Km/H, the reduction was -2.610° (W: 9.631° , W/O: 12.241°) ($p < 0.05$). At 6Km/H, the reduction increased to -3.978° (W: 9.525° , W/O: 13.503°) ($p < 0.05$). These findings indicate that the brace restricts knee internal/external rotation, particularly at higher speeds.

Table 8

Summary Of The Rom Values Of All Joints, Across All Speeds. All Values Are Reported In Degrees

Joints	2Km/H			4Km/H			6Km/H		
	W	W/O	Diff	W	W/O	Diff	W	W/O	Diff
jRight Ankle Abd/Add	7.858	13.983	-6.125	7.996	14.533	-6.538	8.957	16.076	-7.119
jRight Hip Abd/Add	8.579	12.203	-3.623	10.919	14.605	-3.686	13.237	16.263	-3.027
jRight Knee Abd/Add	5.843	6.604	-0.761	5.183	8.558	-3.375	6.087	9.941	-3.853
jRight Ankle Flex/Ext	12.400	21.658	-9.258	18.129	31.340	-13.211	23.307	36.312	-13.006
jRight Hip Flex/Ext	31.286	29.786	1.500	38.468	36.710	1.758	44.012	42.612	1.400
jRight Knee Flex/Ext	51.115	53.349	-2.234	56.866	61.782	-4.916	57.297	62.819	-5.523
jRight Ankle Int/Ext	8.961	15.440	-6.479	10.304	19.319	-9.015	15.186	22.527	-7.340
jRight Hip Int/Ext	12.701	9.317	3.385	12.008	9.627	2.380	11.989	10.550	1.438
jRight Knee Int/Ext	8.790	9.810	-1.020	9.631	12.241	-2.610	9.525	13.503	-3.978

4.2.3 Maximum values results for joint angles. The Right Ankle Abduction/Adduction exhibited consistent reductions in the maximum angles with the brace across all speeds. At 2Km/H, the maximum value decreased significantly from 1.807° without the brace to -1.298° with the brace, a difference of -3.106° ($p < 0.05$). At 4Km/H, the reduction was smaller (-1.036° , W: -1.999° , W/O: 1.845°) ($p < 0.05$). At 6Km/H, the decrease was most pronounced, with a difference of -6.519° (W: -4.039° , W/O: 2.479°) ($p < 0.05$). These findings suggest that the brace heavily restricts ankle abduction/adduction motion, especially at faster speeds.

The Right Hip Abduction/Adduction showed small reductions with the brace at slower and moderate speeds but minimal effects at higher speeds. At 2Km/H, the maximum angle decreased by -1.086° (W: 5.756° , W/O: 6.842°) ($p < 0.05$). At 4Km/H,

the difference was 4.293°, indicating an increase with the brace (W: 6.688°, W/O: 7.829°) ($p < 0.05$). At 6Km/H, the difference was minimal (0.254°, W: 8.560°, W/O: 8.306°) ($p < 0.05$). These results suggest a variable effect of the brace on hip abduction/adduction, with slight restrictions at slower speeds and minimal effects at higher speeds.

The Right Knee Abduction/Adduction displayed consistent reductions in maximum angles with the brace. At 2Km/H, the maximum value decreased from 2.435° to 1.190°, a difference of -1.245° ($p < 0.05$). At 4Km/H, the reduction was -0.540° (W: 1.211°, W/O: 2.903°) ($p < 0.05$). At 6Km/H, the reduction increased to -1.728° (W: 2.045°, W/O: 3.773°) ($p < 0.05$). These findings suggest that the brace consistently restricts knee abduction/adduction motion across all speeds.

The Right Ankle Flexion/Extension exhibited reduced maximum values with the brace across all speeds. At 2Km/H, the maximum value decreased from 12.608° without the brace to 9.845° with the brace, a difference of -2.762° ($p < 0.05$). At 4Km/H, the reduction was -1.912° (W: 9.714°, W/O: 11.626°) ($p < 0.05$). At 6Km/H, the reduction was -1.036° (W: 8.563°, W/O: 9.598°) ($p < 0.05$). These results indicate that the brace limits ankle flexion/extension motion, particularly at slower speeds.

The Right Hip Flexion/Extension showed consistent increases in maximum values with the brace across all speeds. At 2Km/H, the maximum value increased from 21.897° without the brace to 24.537° with the brace, a difference of 2.640° ($p < 0.05$). At 4Km/H, the increase was 3.393° (W: 27.464°, W/O: 24.071°) ($p < 0.05$). At 6Km/H, the brace led to an increase of 4.293° (W: 29.935°, W/O: 25.642°) ($p < 0.05$). These findings suggest that the brace enhances hip flexion/extension motion across all speeds.

The Right Knee Flexion/Extension displayed small reductions in maximum values with the brace across all speeds. At 2Km/H, the difference was minimal (1.440°, W: 54.806°, W/O: 53.365°) ($p < 0.05$). At 4Km/H, the reduction was -1.118° (W: 59.454°, W/O: 60.572°) ($p < 0.05$). At 6Km/H, the difference remained small (-0.540°,

W: 60.229°, W/O: 60.769°) ($p < 0.05$). These results suggest that the brace has minimal impact on knee flexion/extension motion.

The Right Ankle Internal/External Rotation showed reductions in maximum angles with the brace across all speeds. At 2Km/H, the maximum angle decreased from 5.377° to 4.617°, a difference of -0.760° ($p = 0.09$). At 4Km/H, the reduction was more pronounced (-0.844°, W: 5.828°, W/O: 6.672°) ($p < 0.05$). At 6Km/H, the reduction was -2.329° (W: 7.431°, W/O: 9.760°) ($p < 0.05$). These findings suggest that the brace restricts ankle internal/external rotation, particularly at higher speeds.

Table 9

Summary Of The Maximum Values Of The Angles Of All Joints, Across All Speeds. All Values Are Reported In Degrees

Joints	2Km/H			4Km/H			6Km/H		
	W	W/O	Diff	W	W/O	Diff	W	W/O	Diff
jRight Ankle Abd/Add	-1.298	1.807	-3.106	-1.999	1.845	-1.036	-4.039	2.479	-6.519
jRight Hip Abd/Add	5.756	6.842	-1.086	6.688	7.829	4.293	8.560	8.306	0.254
jRight Knee Abd/Add	1.190	2.435	-1.245	1.211	2.903	-0.540	2.045	3.773	-1.728
jRight Ankle Flex/Ext	9.845	12.608	-2.762	9.714	11.626	-1.912	8.563	9.598	-1.036
jRight Hip Flex/Ext	24.537	21.897	2.640	27.464	24.071	3.393	29.935	25.642	4.293
jRight Knee Flex/Ext	54.806	53.365	1.440	59.454	60.572	-1.118	60.229	60.769	-0.540
jRight Ankle Int/Ext	4.617	5.377	-0.760	5.828	6.672	-0.844	7.431	9.760	-2.329
jRight Hip Int/Ext	5.917	3.130	2.787	4.799	3.263	1.536	5.863	4.354	1.509
jRight Knee Int/Ext	4.851	3.737	1.115	5.118	4.510	0.608	4.809	4.669	0.140

The Right Hip Internal/External Rotation demonstrated increased maximum values with the brace across all speeds. At 2Km/H, the maximum angle increased from

3.130° without the brace to 5.917° with the brace, a difference of 2.787° ($p < 0.05$). At 4Km/H, the increase was 1.536° (W: 4.799°, W/O: 3.263°) ($p < 0.05$). At 6Km/H, the brace increased the maximum angle by 1.509° (W: 5.863°, W/O: 4.354°) ($p < 0.05$). These results suggest that the brace enhances hip internal/external rotation across all speeds.

The Right Knee's Internal/External Rotation showed minimal changes with the brace. At 2Km/H, the maximum value increased by 1.115° (W: 4.851°, W/O: 3.737°) ($p < 0.05$). At 4Km/H, the increase was smaller (0.608°, W: 5.118°, W/O: 4.510°) ($p < 0.05$). At 6Km/H, the difference was minimal (0.140°, W: 4.809°, W/O: 4.669°) ($p < 0.05$). These results indicate that the brace has minimal impact on knee internal/external rotation.

4.2.4 Minimum values results for joint angles. The Right Ankle Abduction/Adduction showed small variations in the minimum angles with the brace across all speeds. At 2Km/H, the minimum angle increased slightly with the brace, from -11.498° without the brace to -10.124° with the brace (Diff: 1.373°) ($p < 0.05$). At 4Km/H, the minimum value increased from -12.216° to -10.586°, a difference of 1.630° ($p < 0.05$). At 6Km/H, the difference was smaller (-0.310°, W: -12.346°, W/O: -12.036°) ($p = 0.65$). These results suggest that the brace slightly reduces the range of ankle abduction/adduction, particularly at slower and moderate speeds.

The Right Hip Abduction/Adduction exhibited greater changes with the brace, particularly at faster speeds. At 2Km/H, the minimum angle increased from -5.338° without the brace to -3.729° with the brace, a difference of 1.609° ($p < 0.05$). At 4Km/H, the difference was larger (2.790°, W: -4.054°, W/O: -6.844°) ($p < 0.05$). At 6Km/H, the brace caused a substantial increase of 3.379° (W: -4.835°, W/O: -8.214°) ($p < 0.05$). These findings suggest that the brace reduces the range of hip abduction/adduction consistently, with the largest effects at higher speeds.

The Right Knee Abduction/Adduction showed mixed effects across speeds. At 2Km/H, the minimum angle remained nearly unchanged, with a small difference of -0.026° (W: -3.496°, W/O: -3.470°) ($p < 0.05$). At 4Km/H, the minimum angle increased slightly with the brace by 1.065° (W: -3.426°, W/O: -4.492°) ($p < 0.05$). At 6Km/H, the increase was larger (1.599°, W: -3.635°, W/O: -5.234°) ($p < 0.05$). These

results suggest that the brace slightly reduces knee abduction/adduction range, especially at higher speeds.

The Right Ankle Flexion/Extension exhibited significant changes with the brace across all speeds. At 2Km/H, the minimum angle increased from -9.732° without the brace to -3.390° with the brace, a difference of 6.342° ($p < 0.05$). At 4Km/H, the brace resulted in a large increase of 11.508° (W: -8.897° , W/O: -20.405°) ($p < 0.05$). At 6Km/H, the difference increased further to 14.510° (W: -13.474° , W/O: -27.984°) ($p < 0.05$). These findings highlight that the brace significantly restricts ankle flexion/extension, with the greatest effect observed at higher speeds.

The Right Hip Flexion/Extension displayed small changes with the brace. At 2Km/H, the minimum angle decreased slightly (-0.267° , W: -6.888° , W/O: -6.621°) ($p < 0.05$). At 4Km/H, the difference was minimal (0.343° , W: -11.069° , W/O: -11.412°) ($p < 0.05$). At 6Km/H, the angle slightly increased by 0.928° (W: -13.569° , W/O: -14.497°) ($p < 0.05$). These results suggest that the brace has minimal impact on hip flexion/extension.

The Right Knee Flexion/Extension exhibited greater minimum angles with the brace across all speeds. At 2Km/H, the minimum value increased from 1.132° without the brace to 4.092° with the brace, a difference of 2.960° ($p < 0.05$). At 4Km/H, the increase was 4.048° (W: 3.962° , W/O: -0.085°) ($p < 0.05$). At 6Km/H, the increase was 3.632° (W: 3.210° , W/O: -0.423°) ($p < 0.05$). These results suggest that the brace consistently increases the minimum knee flexion/extension angles, indicating a restricted range.

The Right Ankle Internal/External Rotation exhibited significant increases in minimum angles with the brace. At 2Km/H, the minimum angle increased from -11.044° to -4.075° , a difference of 6.969° ($p < 0.05$). At 4Km/H, the difference was 6.858° (W: -5.164° , W/O: -12.023°) ($p < 0.05$). At 6Km/H, the brace increased the minimum value by 4.651° (W: -7.431° , W/O: -12.081°) ($p < 0.05$). These findings indicate that the brace significantly restricts ankle internal/external rotation across all speeds.

The Right Hip Internal/External Rotation exhibited minimal changes with the brace. At 2Km/H, the minimum angle decreased slightly by -0.625° (W: -6.519° , W/O: -5.894°) ($p < 0.05$). At 4Km/H, the difference was negligible (-0.162° , W: -6.981° , W/O: -6.819°) ($p < 0.05$). At 6Km/H, the brace caused a small reduction of -0.142° (W: -6.469° , W/O: -6.327°) ($p = 0.54$). These findings suggest that the brace has minimal impact on hip internal/external rotation.

The Right Knee Internal/External Rotation demonstrated consistent increases in minimum angles with the brace. At 2Km/H, the minimum angle increased by 1.995° (W: -4.112° , W/O: -6.108°) ($p < 0.05$). At 4Km/H, the increase was 3.123° (W: -4.926° , W/O: -8.049°) ($p < 0.05$). At 6Km/H, the difference was 3.794° (W: -5.032° , W/O: -8.826°) ($p < 0.05$). These findings suggest that the brace consistently restricts the range of knee internal/external rotation.

Table 10

Summary Of The Minimum Values Of The Angles Of All Joints, Across All Speeds. All Values Are Reported In Degrees

Joints	2Km/H			4Km/H			6Km/H		
	W	W/O	Diff	W	W/O	Diff	W	W/O	Diff
jRight Ankle Abd/Add	-10.124	-11.498	1.373	-10.586	-12.216	1.630	-12.346	-12.036	-0.310
jRight Hip Abd/Add	-3.729	-5.338	1.609	-4.054	-6.844	2.790	-4.835	-8.214	3.379
jRight Knee Abd/Add	-3.496	-3.470	-0.026	-3.426	-4.492	1.065	-3.635	-5.234	1.599
jRight Ankle Flex/Ext	-3.390	-9.732	6.342	-8.897	-20.405	11.508	-13.474	-27.984	14.510
jRight Hip Flex/Ext	-6.888	-6.621	-0.267	-11.069	-11.412	0.343	-13.569	-14.497	0.928
jRight Knee Flex/Ext	4.092	1.132	2.960	3.962	-0.085	4.048	3.210	-0.423	3.632
jRight Ankle Int/Ext	-4.075	-11.044	6.969	-5.164	-12.023	6.858	-7.431	-12.081	4.651
jRight Hip Int/Ext	-6.519	-5.894	-0.625	-6.981	-6.819	-0.162	-6.469	-6.327	-0.142
jRight Knee Int/Ext	-4.112	-6.108	1.995	-4.926	-8.049	3.123	-5.032	-8.826	3.794

Chapter 5

Discussions and Conclusions

In this study, we present a comprehensive analysis of muscle activity and joint kinematics in the context of wearing a knee-ankle-foot orthosis (KAFO) brace during gait. Root mean square (RMS), peak muscle amplitudes, coefficient of variability (CoV), maximum and minimum joint angles, and range of motion (ROM) are all covered in the results, over a wide range of parameters. This work provides biomechanical insights into the adaptations induced by the brace and their implications for stability, energy efficiency, and muscular activation.

5.1 Muscle Activation Patterns

5.1.1 Biceps Femoris. The Biceps Femoris showed a consistent increase in activation at all speeds with the brace. This is evidenced by increased peak amplitudes, and higher RMS and MAV values during the loading response phase. The results indicate that the Biceps Femoris plays the role of compensating for the mechanical restrictions of the brace, particularly at slower speeds, where the differences are more pronounced. Yet, the decrease in effect at higher speeds suggests that some adaptation or reliance on other muscular strategies to maintain gait efficiency might be occurring.

This increased activation of the Biceps Femoris is interpreted as a mechanism to maintain knee stability throughout the gait cycle. This pattern is consistent with previous studies showing the Biceps Femoris to be a posterior knee supporter. Furthermore, the elevated activation levels during the loading response phase indicate the importance of this muscle in regulating knee flexion during braced conditions. These patterns may, over time, affect muscular fatigue and deserve further investigation.

5.1.2 Gastrocnemius Lateralis and Medialis. On the other hand, the Gastrocnemius Lateralis showed decreased activation through most conditions, especially at lower speeds. The brace may reduce this demand by providing ankle stabilization, thereby reducing the need for the gastrocnemius muscles to compensate for joint instability. Meanwhile, the Gastrocnemius Medialis had minimal changes, indicating a lesser role in adapting to the braced condition. These observations suggest that wearing the brace shifts the muscular workload from distal to proximal regions of the leg.

The Gastrocnemius Lateralis activation reduction observed demonstrates that the brace is capable of providing external stability at the ankle joint. This could correspond to energy savings during prolonged walking as a result of the reduction in muscle activity. Yet the minimal effect on the Gastrocnemius Medialis indicates that not all muscles in the posterior compartment react equally to bracing. The finding may have implications for designing braces that are best suited to particular muscular strategies.

5.1.3 Vastus Lateralis and Medialis. The brace increased activation of both Vastus muscles, predominantly at slower and moderate speeds. It is likely these muscles play a role in helping to stabilize the knee joint and to maintain alignment. The observed reductions in variability (CoV) of Vastus Lateralis activation are consistent with increased activation in this muscle during gait.

The level of activation in Vastus muscles is elevated with their known role as primary knee extensors aiding to provide stability during the stance phase. Further increased recruitment of these muscles by the brace to counteract the limited knee mobility probably did not occur due to the mechanical restrictions placed by the brace. Interestingly, the reduction in variability is greatest for the Vastus Lateralis, suggesting that the improved neuromuscular control may be beneficial for those with impaired coordination or balance.

5.1.4 Tibialis Anterior. RMS and MAV values for the Tibialis Anterior consistently showed that the brace reduced activation. The mechanical assistance of the brace in dorsiflexion may have reduced the load on this muscle, leading to this reduction. The brace was found to stabilize the ankle and decrease muscular effort across all speeds, a consistent pattern.

The Tibialis Anterior activation may be reduced because the brace assists in foot clearance during the swing phase. The risk of tripping or stumbling could be mitigated by this assistance, especially for those with dorsiflexion weakness. Nevertheless, adjunctive strengthening exercises are important as long-term reliance on the brace may cause muscular atrophy in the Tibialis Anterior.

5.1.5 Variability in Muscle Activation (CoV). CoV analysis of the brace indicated that on average, the brace decreased variability in muscle activation of key muscles like Biceps Femoris, Gastrocnemius Lateralis, and Gastrocnemius Medialis. Greater stability is sustained, with reduced variability representing the maintenance of more consistent and controlled muscle activation patterns during gait. In addition, the brace can help minimize fluctuations in muscle activity, thus being highly helpful for people with gait abnormalities and instability. End effector variability decreased (greater neuromuscular coordination and predictability based on the muscular engagement during gait cycles). The effect is most pronounced at moderate speeds where variability peaks under unbraced conditions. Hopefully, the stabilizing effect of the brace will reduce the amount of cognitive load needed to maintain balance and therefore make the gait pattern smoother and more efficient.

5.2 Joint Kinematics

5.2.1 Ankle Joint. Ankle abduction/adduction and flexion/extension were significantly restricted by the brace across all speeds. At higher speeds, the brace was restrictive and the most pronounced reductions were seen. However, this limitation can increase joint stability, but may also limit natural gait patterns, resulting in increased energy expenditure. A trade-off between stability and mobility is seen in the restricted motion at the ankle. The reduced ROM reduces the risk of joint instability but does change gait biomechanics shifting the workload to more proximal joints like the knee and hip. If sustained over long periods, these adjustments could result in overuse injuries.

5.2.2 Knee Joint. RMS values of knee flexion/extension slightly increased at slower speeds, indicating that the brace enables controlled knee movement. However, reductions in ROM suggest the brace constrains knee mobility, which may impact stride length and overall gait efficiency. The complex interplay between stabilization and mobility is reflected in the mixed effects on knee abduction/adduction and internal/external rotation.

This dual role of the brace in increasing stability and limiting excessive movement is further evidenced by the effects of the brace on knee kinematics. Reduced abduction/adduction variability observed in the current results indicates improved lateral control which is important for those with medial or lateral instability.

Nevertheless, the limited ROM in flexion/extension may require the use of strategies at other joints to continue forward progression.

5.2.3 Hip Joint. Also, small increases in ROM and RMS values were observed in hip flexion/extension with the brace, indicating that this joint became more mobile. The motion increase at the hip may be a mechanism to balance the restrictions at the knee and ankle. However, ROM in hip abduction/adduction was consistently reduced, especially at moderate speeds, suggesting the effect of the brace on lateral stability.

This corresponds to prior research that shows proximal compensation in response to distal joint restrictions. This adaptation preserves stride length and overall gait progression in the face of the constraints imposed by the brace. This may reduce lateral motion at the hip, however, and limit manoeuvrability, especially during activities where quick directional changes are needed.

5.3 Implications for Gait Stability and Efficiency

The dual role of the brace in providing stability and imposing mechanical constraints is highlighted by the impact of the brace on muscle activation and joint kinematics. Reduced variability and increased activation of stabilizing muscles (e.g., Biceps Femoris, Vastus Lateralis) improve gait control, but the restricted ROM and altered joint angles may decrease gait efficiency. Such findings suggest the necessity to customize the fine-tuning between stability and natural movement patterns.

In particular, the observed improvements in stability are particularly relevant for populations with neuromuscular disorders or joint instability. The brace provides a viable method for mitigating fall risk and enhancing confidence during ambulation through enhanced muscular control and reduced variability. While the increased energy expenditure associated with higher energy expenditure, may cause fatigue during prolonged use, the mechanical constraints on joint motion may force higher energy expenditure.

5.4 Speed-Dependent Effects

Distinct speed-dependent effects of the brace are revealed by the results. Increased muscle activation and decreased variability at slower speeds suggest increased reliance on mechanisms. With increased speed, the brace significantly reduces the ROM of the brace and may limit stride efficiency. These observations

emphasize that brace design and usage should be tailored to a given gait speed and functional demand.

The stabilizing influence of the brace is most apparent at slower speeds as reflected in increased control and reduced variability. At higher paces, the mechanical capacity of the brace restricts natural gait dynamics, and more effort is needed to keep up with the same pace. This trade-off demonstrates the necessity of dynamic bracing solutions that can adjust to changing functional demands

5.5 Clinical Implications

The findings are useful for people with neuromuscular disorders or joint instability to understand the therapeutic benefits and limitations of KAFO braces. The increased activation of stabilizing muscles and decreased variability indicate that braces may be useful in the enhancement of gait control. However, caution must be exercised in assessing the restrictive effects on ROM and potential energy costs.

In addition to stabilizing effects, KAFO braces have clinical utility. Braces promote consistent muscle activation patterns and minimize variability making them useful for rehabilitation because they improve neuromuscular coordination. However, braces may be prescribed for long periods of time for treating functional problems, but clinicians must weigh the downside: restricted joint mobility and risk of muscle atrophy.

5.6 Future Directions

Long-term adaptations to brace usage, such as muscular strength, endurance, and joint mechanics, should be further researched. This inherently limits new technologies in resistance bracing, like real-time biofeedback and dynamic changes, which could increase the functionality of a brace and create a more natural and efficient gait. Furthermore, studies assessing the psychological and functional outcomes of wearing a brace would provide a more comprehensive picture of its effect on quality of life. Future investigations should also include the role of individualized bracing strategies in optimizing outcomes. The ability of clinicians to apply brace design to specific biomechanical needs permits a maximization of therapeutic benefit and minimization of adverse effect. Moreover, wearable sensors and machine learning algorithms can be integrated to provide real-time monitoring and adjustments to optimize brace efficacy.

5.7 Conclusion

This study finds that stabilization and mobility in individuals wearing KAFO braces are a complex interaction. This research contributes to the optimization of brace design and therapeutic strategies for improving gait stability and efficiency by elucidating the biomechanical effects on muscle activity and joint kinematics. The results demonstrate the need for individualized approaches to bracing to maximize its benefits while minimizing its costs.

KAFO braces promote stability and decrease variability in terms of gait. But because of their ability to restrict joint mobility, a delicate balance between stability and natural movement patterns is absolutely required. This study's results show that although the brace reduces variability and increases the activation of stabilizing muscles, it also restricts the range of motion at the ankle and knee joints. The trade-off between these factors constitutes a challenge that clinicians should weigh appropriately to identify the optimal outcomes for the different patient populations.

Additionally, the speed-dependent effects demonstrated highlight the need for functional assessment at multiple gait speeds when designing and prescribing braces. More flexible solutions for users could be provided by the integration of advanced technologies such as dynamic bracing systems that respond to real-time biomechanical demands. The introduction of these innovations could not only improve a person's efficiency of gait but could also improve the quality of his or her life in general for individuals that rely on KAFO braces.

Clinicians and researchers need to work together to address the long-term implications of brace use (muscular deconditioning or joint stiffness). Future interventions will seek to overcome these challenges through the use of multidisciplinary approaches and leverage the stability benefits delivered by the brace. Overall, the presented findings in this paper open doors to future custom and improved bracing solutions that lead to better mobility and independence in people with gait impairments.

5.8 Limitation

There is significant value in this research to better understand the impact of KAFO braces on the biomechanics of gait and muscle force production and joint trajectories. Nevertheless, the following are five limitations that need to be understood, given their impact on the generalization of the presented study results and the definition of further research directions. These limitations cut across the participants, the study and the experimental procedure, and even the general use of braces.

1. Participant Characteristics, and Number of Participants

Most of the subjects were relatively high-functioning and the questions raised pertained more to low functioning, patients with severe neuromuscular conditions, as well as patients with other demographic differences in terms of age or physical size (BMI, height, etc).

2. Short-Term Focus

No studies included long-term dynamics of physical well-being and what happens to muscles or other parts of the body when not used regularly in the long term.

3. Comfort of brace and how it fits to the wearer

This was important because the type of brace used did not consider the bracing adjustments that sometimes patients need in a clinical setting.

4. Issues that relate to the risk of overuse injuries

strategies increase the risk of overuse injuries which we did not consider in our study but are crucial for chronic use.

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