

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

**INVESTIGATING THE EFFICACY OF A 3D-PRINTED HIP BRACE
ACTUATION SYSTEM IN MINIMIZING LOWER-LIMB MUSCLE
ACTIVATION DURING SQUATTING**

MASTER'S THESIS

MOHAMMED IMAD TARRAD ALGRAITI

ISTANBUL 2025

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

**INVESTIGATING THE EFFICACY OF A 3D-PRINTED HIP BRACE
ACTUATION SYSTEM IN MINIMIZING LOWER-LIMB MUSCLE
ACTIVATION DURING SQUATTING**

MASTER'S THESIS
MOHAMMED IMAD TARRAD ALGRAITI

THESIS ADVISOR
ASST. PROF. BORA BÜYÜKSARAÇ

ISTANBUL 2025



T.C.
BAHCESEHIR UNIVERSITY
GRADUATE SCHOOL

29/05/2025

MASTER THESIS APPROVAL FORM

Program Name:	BIOMEDICAL ENGINEERING (ENGLISH, THESIS)
Student's Name and Surname:	Mohammed Imad Tarrad Algraiti
Name of the Thesis:	Investigating the Efficacy of a 3D-Printed Hip Brace Actuation System in Minimizing Lower-Limb Muscle Activation During Squatting
Thesis Defense Date:	29/05/2025

This thesis has been approved by the Graduate School which has fulfilled the necessary conditions as Master thesis.

Assoc. Prof. Yücel Batu SALMAN
Director of Graduate School

This thesis was read by us, quality and content as a Master's thesis has been seen and accepted as sufficient.

	Title, Name	Institution	Signature
Thesis Advisor	Asst. Prof. Bora Büyüksaraç	Bahçeşehir University	
2sd Member	Asst. Prof. Firat Matur	Bahçeşehir University	
3rd Member (Outside Institution)	Asst. Prof. Mustafa Erkam Özateş	Turkish-German University	



I hereby declare that all information in this document has been obtained and presented in accordance with academic rules and ethical conduct. I also declare that, as required by these rules and conduct, I have fully cited and referenced all material and results that are not original to this work.

Name, Last Name: Mohammed
Imad Tarrad Algraiti

Signature:

ABSTRACT

INVESTIGATING THE EFFICACY OF A 3D-PRINTED HIP BRACE ACTUATION SYSTEM IN MINIMIZING LOWER-LIMB MUSCLE ACTIVATION DURING SQUATTING

Mohammed Imad Tarrad Algraiti
Master's Program in Biomedical Engineering
Supervisor: Asst. Prof. Bora Büyüksaraç

May 2025, 99 Pages

Musculoskeletal injuries to the hip and lower back remain a major concern in occupations involving repetitive lifting or squatting. This study introduces a hip exoskeleton brace equipped with a 3D-printed actuator, designed to reduce muscle effort specifically during the ascending phase of bodyweight squats, where mechanical demand on the hip extensors is greatest. The actuator, fixed onto the brace and operating via a 9:1 gear ratio, provided targeted torque assistance without external load.

To evaluate the biomechanical impact of the brace, surface electromyography (sEMG) and motion capture systems were employed. Muscle activity was recorded from five hip-related muscles—Biceps Femoris, Gluteus Maximus, Gluteus Medius, Rectus Femoris, and Tensor Fasciae Latae—using eight time-domain EMG features: Root Mean Square, Mean Absolute Value, Area Under the Curve, Variance. Kinematic data were extracted from the right hip, knee, and ankle joints using RMS, MAV, and Range of Motion to assess joint-level mechanical behavior during assistance.

Significant reductions in muscle activation were observed in Rectus Femoris and Gluteus Medius, indicating effective muscular offloading during the supported

phase of squatting. Additionally. From a kinematic perspective, hip flexion/extension and internal/external rotation showed significant reductions, while hip abduction/adduction increased notably under brace support. At the knee, changes in flexion/extension, internal/external rotation, and abduction/adduction further reflected joint-level adaptations.

These results confirm that the actuator-brace system effectively offloads specific hip musculature during assisted squatting. Joint-level changes at the hip and knee reflect mechanical redistribution under brace support, supporting its potential as a targeted ergonomic intervention in repetitive lifting scenarios.

Keywords: Hip Exoskeleton Brace, 3D-Printed Actuator, Squat Biomechanics, Muscle Offloading, Wearable Robotics

ÖZET

ÇÖMELME SIRASINDA ALT EKSTREMİTE KAS AKTİVASYONUNU EN AZA İNDİRMEDE 3D BASKILI KALÇA DESTEĞİ ÇALIŞTIRMA SİSTEMİNİN ETKİNLİĞİNİN ARAŞTIRILMASI

Mohammed Imad Tarrad Algraiti

Biyomedikal Mühendisliği Yüksek Lisans Programı

Tez Danışmanı: Dr. Öğr. Üyesi Bora Büyüksaraç

Mayıs 2025, 99 sayfa

Kalça ve belde kas-iskelet yaralanmaları, tekrarlayan kaldırma veya çömelme içeren mesleklerde önemli bir endişe kaynağı olmaya devam etmektedir. Bu çalışmada, özellikle kalça ekstansörleri üzerindeki mekanik talebin en yüksek olduğu vücut ağırlığı çömelmelerinin yükselme aşamasında kas eforunu azaltmak için tasarlanmış, 3D baskılı bir aktüatörle donatılmış bir kalça dış iskelet desteği tanıtılmaktadır. Destek üzerine sabitlenen ve 9:1 dişli oranıyla çalışan aktüatör, harici yük olmadan hedeflenen tork yardımı sağladı.

Korsenin biyomekanik etkisini değerlendirmek için yüzey elektromiyografisi (sEMG) ve hareket yakalama sistemleri kullanıldı. Kas aktivitesi, sekiz zaman alanı EMG özelliği kullanılarak kalçayla ilgili beş kastan (Biceps Femoris, Gluteus Maximus, Gluteus Medius, Rectus Femoris ve Tensor Fasciae Latae) kaydedilmiştir: Kök Ortalama Kare, Ortalama Mutlak Değer, Eğri Altındaki Alan, Varyans. Yardım sırasında eklem düzeyindeki mekanik davranışı değerlendirmek için RMS, MAV ve Hareket Aralığı kullanılarak sağ kalça, diz ve ayak bileği eklemlerinden kinematik veriler çıkarılmıştır.

Rectus Femoris ve Gluteus Medius'ta kas aktivasyonunda önemli azalmalar gözlenmiştir, bu da çömelmenin desteklenen fazı sırasında etkili kas yüklemesini

göstermektedir. Ek olarak. Kinematik açıdan bakıldığında, kalça fleksiyonu/ekstansiyonu ve iç/dış rotasyonu önemli ölçüde azalma gösterirken, kalça abdüksiyonu/addüksiyonu destek altında önemli ölçüde artmıştır. Dizde fleksiyon/ekstansiyon, iç/dış rotasyon ve abdüksiyon/addüksiyondaki değişiklikler eklem düzeyindeki adaptasyonları daha da yansıtmıştır.

Bu sonuçlar, aktüatör-korse sisteminin destekli çömelme sırasında belirli kalça kaslarına etkili bir şekilde yük bindirdiğini doğrulamaktadır. Kalça ve dizdeki eklem seviyesi değişiklikleri, korse desteği altında mekanik yeniden dağılımı yansıtmakta ve tekrarlayan kaldırma senaryolarında hedeflenen ergonomik bir müdahale olarak potansiyelini desteklemektedir.

Anahtar Kelimeler: Kalça Dış Iskelet Desteği, 3D Baskılı Aktüatör, Çömelme Biyomekaniği, Kas Boşaltma, Giyilebilir Robotik

ACKNOWLEDGEMENTS

I would like to express my deepest gratitude to my advisor, Asst. Prof. Bora Büyüksaraç, for his unwavering guidance, support, and mentorship throughout this research. His trust in my abilities and consistent encouragement challenged me to push beyond my limits and achieve more than I ever thought possible.



TABLE OF CONTENTS

ETHICAL CONDUCT	iii
ABSTRACT	iv
ÖZET.....	vi
ACKNOWLEDGEMENTS	viii
TABLE OF CONTENTS	IX
LIST OF TABLES	XIII
LIST OF FIGURES.....	XV
LIST OF ABBREVIATIONS	XVIII
1 Chapter 1 Introduction	1
1.1 Background.....	1
1.2 Purpose Of The Study	2
1.3 Research Questions And Hypotheses	3
1.4 Significance Of The Study	3
2 Chapter 2 Literature Review	5
2.1 Biomechanics Of Bodyweight Squatting And Exoskeletons For Assistance And Injury Prevention	5
2.2 Functional Roles Of Key Muscles During Squatting	8
2.2.1 Biceps Femoris (Hamstrings).....	9
2.2.2 Gluteus Maximus.	10
2.2.3 Gluteus Medius.	11
2.2.4 Rectus Femoris.....	13
2.2.5 Tensor Fasciae Latae (Tfl).	14
2.3 Actuator Technologies In Exoskeletons And The Role Of Gear Ratio (9:1) In Performance And Safety.....	16

2.3.1	Electric Actuators (Motors And Gear Drives).	17
2.3.2	Pneumatic Actuators (Artificial Muscles And Air Cylinders).	21
2.3.3	Hydraulic Actuators.	24
2.3.4	Gear Ratio Considerations Across Technologies.	27
2.4	Surface Emg Signal Analysis In Human Movement Studies	29
2.5	Motion Capture In Squat Biomechanics And Kinematic Feature Analysis	31
2.6	Synthesis Of Literature And Current Gaps.....	36
3	Chapter 3 Materials And Methods.....	39
3.1	Participant Selection And Study Design.	39
3.2	Instrumentation And Devices	39
3.2.1	Electromyography (EMG).	39
3.2.2	Motion Capture (Xsens).	40
3.2.3	Powered Hip Support Brace.....	41
3.3	Data Acquisition.	42
3.3.1	Electromyography (EMG).	42
3.3.2	Motion Capture (Xsens).	43
3.4	Exoskeleton System Design.	44
3.4.1	Hip Brace And Actuator Design.	44
3.4.2	Control And Sensing.	46
3.5	Experimental Protocol.	48
3.6	Data Analysis.....	49
3.7	Statistical Analysis.	51
4	Chapter 4 Results	53
4.1	Full Squat Cycle Analysis	53
4.1.1	Visual Overview Of The Time-Normalized Data Profiles.....	53

4.1.2	Electromyography (sEMG) Results – Full Squat.....	54
4.1.3	Kinematic Results – Full Squat.....	57
4.1.4	Kinematic Correlation Analysis – Full Squat.	61
4.2	Descent Phase Analysis.....	62
4.2.1	Electromyography (sEMG) Results – Descent Phase.....	62
4.2.2	Kinematic Results – Descent Phase.	64
4.3	Ascent Phase Analysis.....	68
4.3.1	Electromyography (sEMG) Results – Ascent Phase.....	68
4.3.2	Kinematic Results – Ascent Phase.....	71
4.3.3	Kinematic Correlation Analysis – Ascent Phase.....	75
4.4	Summary Of Key Findings.....	76
5	Chapter 5 Discussion	80
5.1	Overview Of Principal Findings.....	80
5.2	Interpretation Of Findings During The Ascent Phase	81
5.2.1	Effects On Muscle Activity (sEMG) In Relation To Brace Assistance.....	81
5.2.2	Kinematic Alterations And Their Implications For Ascent Performance..	83
5.2.3	Hip-Knee Kinematic Coupling During Ascent.	86
5.2.4	Synthesis Of Hip-Knee Coupling During Ascent.	88
5.3	Interpretation Of Findings During The Full Squat Cycle And Descent Phase..	89
5.4	Synthesis Of Findings: Efficacy, Unexpected Outcomes, And Design Implications	92
5.4.1	Efficacy In Assisting Ascent And Modulating Muscle Activity.....	92
5.4.2	Lack Of Significant Semg Alterations In Gluteus Maximus, Biceps Femoris, And Tensor Fasciae Latae.....	93
5.4.3	Brace-Induced Alterations In Hip, Knee, And Ankle Non-Sagittal Plane Movements.....	94

5.4.4	Non-Sagittal Kinematic Alterations Raise Several Critical Points	94
5.4.5	Influence Of Passive Brace Mechanics.	96
6	Chapter 6 General Conclusion, Limitations & Future Work.....	97
6.1	General Conclusion	97
6.2	Limitations & Future Work	98
7	REFERENCES	100



LIST OF TABLES

TABLES

Table 1 Descriptive Statistics Mean (SD) And Statistical Comparison (P-Values) Of Sema Features (RMS, MAV, AUC, VAR) For Selected Lower Limb Muscles During The Full Squat Cycle Performed With And Without The Hip Brace.	54
Table 2 Percentage Change (%) In Sema Features For Selected Lower Limb Muscles When Wearing The Hip Brace Compared To The No-Brace Condition During The Full Squat Cycle.	55
Table 3 Descriptive Statistics Mean (SD) And Statistical Comparison (P-Values) Of Kinematic Features (ROM, RMS, MAV) For Right Hip, Knee, And Ankle Joint Movements During The Full Squat Cycle Performed With And Without The Hip Brace.	57
Table 4 Percentage Change (%) In Kinematic Features For Right Hip, Knee, And Ankle Joint Movements When Wearing The Hip Brace Compared To The No-Brace Condition During The Full Squat Cycle.	58
Table 5 Spearman's Rank Correlation Coefficients (P) And P-Values For The Relationship Between Brace-Induced Change Scores (Δ) In Hip Flexion/Extension Features And Knee Flexion/Extension Features During The Full Squat Cycle.....	61
Table 6 Descriptive Statistics Mean (SD) And Statistical Comparison (P-Values) Of Sema Features (RMS, MAV, AUC, VAR) For Selected Lower Limb Muscles During The Descent Phase Of The Squat Performed With And Without The Hip Brace.....	62
Table 7 Descriptive Statistics Mean (SD) And Statistical Comparison (P-Values) Of Kinematic Features (ROM, RMS, MAV) For Right Hip, Knee, And Ankle Joint Movements During The Descent Phase Of The Squat Performed With And Without The Hip Brace.....	64
Table 8 Percentage Change (%) In Kinematic Features For Right Hip, Knee, And Ankle Joint Movements When Wearing The Hip Brace Compared To The No-Brace Condition During The Descent Phase Of The Squat. Only Statistically Significant ($P < 0.05$) Percentage Changes.....	65

Table 9 Descriptive Statistics Mean (SD) And Statistical Comparison (P-Values) Of Seng Features (RMS, MAV, AUC, VAR) For Selected Lower Limb Muscles During The Ascent Phase Of The Squat Performed With And Without The Hip Brace.	68
Table 10 Percentage Change (%) In Seng Features For Selected Lower Limb Muscles When Wearing The Hip Brace Compared To The No-Brace Condition During The Ascent Phase Of The Squat. Only Statistically Significant ($P < 0.05$) Percentage Changes Are Listed; Values Are Rounded To The Nearest Whole Number.....	69
Table 11 Descriptive Statistics Mean (SD) And Statistical Comparison (P-Values) Of Kinematic Features (ROM, RMS, MAV) For Right Hip, Knee, And Ankle Joint Movements During The Ascent Phase Of The Squat Performed With And Without The Hip Brace.....	72
Table 12 Percentage Change (%) In Kinematic Features For Right Hip, Knee, And Ankle Joint Movements When Wearing The Hip Brace Compared To The No-Brace Condition During The Ascent Phase Of The Squat.	73
Table 13 Spearman's Rank Correlation Coefficients (P) And P-Values For The Relationship Between Brace-Induced Change Scores (Δ) In Hip Flexion/Extension Features And Knee Flexion/Extension Features During The Ascent Phase Of The Squat.	76

LIST OF FIGURES

FIGURES

Figure 1 A Diagram Illustrating The Squat-Assist Design Used In Industrial Leg-Support Exoskeletons (Kuber Et Al., 2023).	1
Figure 2 Anterior And Posterior Views Of The Leg Anatomy (Marzullo, 2022).	8
Figure 3 From Left To Right — (A) Motor Sub-Assembly, (B) Gearbox Sub-Assembly, And (C) Cross-Sectional Cad View Of The Complete Actuator Assembly, Designed To Fit Within Dimensions Of 126×68 Mm (Singh Et Al., 2020).	17
Figure 4 Illustrates A Planetary Gear System (Also Known As An Epicyclic Gear Train), Which Consists Of A Central Sun Gear, Multiple Planet Gears, A Surrounding Ring Gear With Internal Teeth, And A Planet Carrier That Holds And Supports The Planet Gears.	19
Figure 5 Pneumatic Artificial Muscles Exoskeletal Design (Zhong Et Al., 2020).	21
Figure 6 Cross-Sectional View Of A Swash Plate Axial Piston Pump. Exemplifies How Hydraulic Systems Can Efficiently Convert Fluid Power Into Linear Or Rotary Motion In Compact Actuator Designs (Sun Et Al., 2021).	24
Figure 7 Trigno Wireless Sema System (Gage, 2023).	40
Figure 8 Trigno® Wireless Biofeedback (SN: SP-W02C-1357) System (Gage, 2020). ..	41
Figure 9 The Powered Hip Support Brace	41
Figure 10 Lower Limb EMG Sensor Placement By SENIAM.	42
Figure 11 Movella Xsens Motion Capture System Sensor Placement.	43
Figure 12 Actuator Design Inspired By The Open-Source Opentorque Actuator.	45
Figure 13 Eaglepower 8308 Brushless Outrunner.	46
Figure 14 Brushless DC Motor Driver (Model ZS-X11F).	47
Figure 15 Mean ($\pm$ Standard Deviation) Time/Amplitude Normalized Sema Amplitude Profiles For Gluteus Maximus, Gluteus Medius, Rectus Femoris, Tensor Fasciae Latae, And Biceps Femoris During The Full Squat Cycle. Data Are Shown For Conditions With The Hip Brace (Blue Lines And Shading) And Without The Hip Brace (Red Lines And Shading).	53

Figure 16 Mean ($\pm$ Standard Deviation) Time-Normalized Joint Angle Profiles. Data Are Shown For Conditions With The Hip Brace (Blue Lines And Shading) And Without The Hip Brace (Red Lines And Shading).....	54
Figure 17 Percentage Change In Sema Features (RMS, MAV, AUC) For The Rectus Femoris Muscle During The Full Squat Cycle When Using The Hip Brace. The Bars Represent The Percentage Change In EMG Features When Comparing The ‘With Brace’ Condition To The ‘Without Brace’ (Baseline) Condition. Negative Values Indicate A Reduction In Muscle Activity With The Brace For These Features.	56
Figure 18 Heatmap Illustrating The Percentage Change In Kinematic Features (ROM, MAV, RMS) Across Hip, Knee, And Ankle Joint Movements During The Full Squat Cycle When Comparing The 'With Brace' To The 'Without Brace' (Baseline) Conditions. Color Intensity Indicate The Magnitude And Direction Of Change (Warmer Colors For Increases, Cooler Colors For Decreases). Ns Indicates Non-Significant Changes.	59
Figure 19 Scatter Plots Illustrating Significant Positive Correlations Between Brace-Induced Change Scores (Δ) In Selected Hip Flexion/Extension (FE) Features And Knee Flexion/Extension (FE) Features During The Full Squat Cycle. (A) Δ Hip ROM Vs. Δ Knee RMS, (B) Δ Hip ROM Vs. Δ Knee MAV, (C) Δ Hip RMS Vs. Δ Knee RMS, (D) Δ Hip RMS Vs. Δ Knee MAV. Spearman's Rho (ρ) And P-Values Are Shown For Each Plot.	62
Figure 20 Heatmap Illustrating The Percentage Change In Kinematic Features (ROM, MAV, RMS) Across Hip, Knee, And Ankle Joint Movements During The Descent Phase Of The Squat When Comparing The 'With Brace' To The 'Without Brace' (Baseline) Conditions. Color Intensity Indicate The Magnitude And Direction Of Change (Warmer Colors For Increases, Cooler Colors For Decreases). NS Indicates Non-Significant Changes.	66
Figure 21 Percentage Change In Sema Features (RMS, MAV, AUC) For The Gluteus Medius Muscle During The Ascent Phase Of The Squat When Using The Hip Brace. Bars Represent The Percentage Change When Comparing The 'With Brace' Condition To The 'Without Brace' (Baseline) Condition. Negative Values Indicate A Reduction In Muscle Activity.....	70

Figure 22 Percentage Change In Sema Features (RMS, MAV, AUC) For The Rectus Femoris Muscle During The Ascent Phase Of The Squat When Using The Hip Brace. Bars Represent The Percentage Change When Comparing The 'With Brace' Condition To The 'Without Brace' (Baseline) Condition. Negative Values Indicate A Reduction In Muscle Activity.....	71
Figure 23 Heatmap Illustrating The Percentage Change In Kinematic Features (ROM, MAV, RMS) Across Hip, Knee, And Ankle Joint Movements During The Ascent Phase Of The Squat When Comparing The 'With Brace' To The 'Without Brace' (Baseline) Conditions. Color Intensity Indicate The Magnitude And Direction Of Change (Warmer Colors For Increases, Cooler Colors For Decreases). NS Indicates Non-Significant Changes.	73
Figure 24 Scatter Plots Illustrating Significant Positive Correlations Between Brace-Induced Change Scores (Δ) In Selected Hip Flexion/Extension (FE) Features And Knee Flexion/Extension (FE) Features During The Ascent Phase Of The Squat. (A) Δ Hip ROM Vs. Δ Knee RMS, (B) Δ Hip RMS Vs. Δ Knee RMS, (C) Δ Hip RMS Vs. Δ Knee MAV. Spearman's Rho (ρ) And P-Values Are Shown For Each Plot.	76
Figure 25 Summary Of Significant Percentage Changes In Sema And Kinematic Features During The Full Squat And Ascent Phases When Wearing The Hip Brace. Bubble Size Represents The Absolute Magnitude Of The Percent Change, And Color Indicates The Direction And Magnitude Of Change (Cooler Colors For Decreases, Warmer For Increases). Cells Marked “NS” Denote Non-Significant Changes ($P \geq 0.05$).	77

LIST OF ABBREVIATIONS

ABS	Acrylonitrile Butadiene Styrene
AUC	Area Under the Curve
BF	Biceps Femoris
BLEEX	Berkeley Lower Extremity Exoskeleton
BLDC	Brushless DC (motor)
CAD	Computer-Aided Design
DC	Direct Current
DOF	Degrees of Freedom
EHA	Electro-Hydrostatic Actuator
EMG	Electromyography
HMI	Human-Machine Interface
IMU	Inertial Measurement Unit
IT	Iliotibial (band)
MAV	Mean Absolute Value
MVC	Maximum Voluntary Contraction
PAM	Pneumatic Artificial Muscle
PCA	Principal Component Analysis
PWM	Pulse Width Modulation
QDD	Quasi-Direct Drive
RF	Rectus Femoris
RMS	Root Mean Square
ROM	Range of Motion
SD	Standard Deviation
SEAs	Series Elastic Actuators
TFL	Tensor Fasciae Latae
WMSD	Work-Related Musculoskeletal Disorder

Chapter 1

Introduction

1.1 Background

Musculoskeletal injuries affect the hip and lower back are common in tasks involving repetitive squatting, particularly in occupational and rehabilitation settings. These injuries often develop gradually due to the repeated mechanical loading of joints and muscles during daily movements, even in the absence of external weights (Gallagher & Marras, 2012). While squatting is considered biomechanically safer than stooping for lifting, it places high demands on the hip and knee extensors, increasing the risk of overuse and fatigue-related injuries over time (Escamilla, 2001).

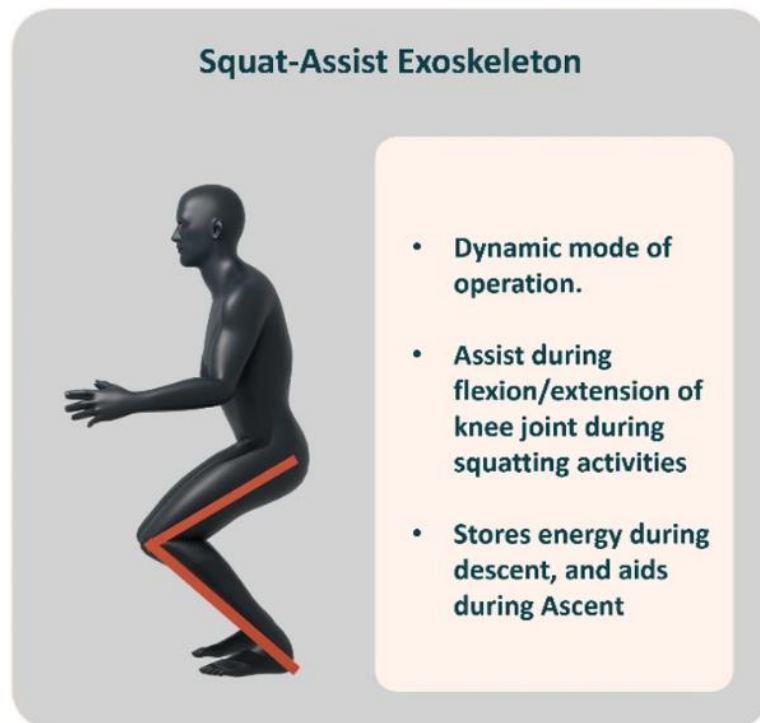


Figure 1. A diagram illustrating the squat-assist design used in industrial leg-support exoskeletons (Kuber et al., 2023).

The cumulative strain from repeated squatting movements can compromise joint stability and muscle endurance, especially when performed frequently across long durations. Although proper technique can help distribute load more effectively, degradation caused by fatigue often leads to compensatory patterns that elevate the risk of

tissue stress and injury (Parijat & Lockhart, 2008). Therefore, strategies aimed at supporting the body during squat-related movements are gaining attention, particularly in contexts where movement repetition is unavoidable.

Wearable assistive devices such as exoskeletons have emerged as potential solutions to alleviate biomechanical stress. These systems are designed to assist movement, reduce muscle activation, and improve joint efficiency (de Looze et al., 2016). However, many current designs remain bulky, costly, or lack precision in supporting the complex mechanics of the hip joint (Junius et al., 2017).

To address these limitations, this study explores the use of a hip exoskeleton system featuring a lightweight, 3D-printed actuator mounted on the rotational axis of a commercial orthopedic hip brace. The device was developed to assist only during the ascending phase of the squat, where mechanical demand is highest. The aim is to investigate whether targeted assistance can reduce muscle effort and positively influence joint behavior in healthy individuals performing bodyweight squats.

1.2 Purpose Of The Study

The purpose of this study is to evaluate the effectiveness of a hip exoskeleton brace equipped with a lightweight, 3D-printed actuator in reducing muscle activation and influencing joint kinematics during unweighted squatting tasks. The exoskeleton is designed to assist the hip joint specifically during the ascending phase of the squat, where mechanical demand on the hip extensors is greatest.

By comparing electromyographic and kinematic data between assisted and unassisted squats in healthy adult males, the study aims to determine whether the actuator-brace system can provide meaningful muscular offloading without disrupting natural movement patterns. The findings are intended to inform the development of ergonomic, low-cost assistive devices for injury prevention in repetitive, lower-body movement tasks.

1.3 Research Questions And Hypotheses

Research Questions

- Can the application of a hip exoskeleton brace with a lightweight actuator reduce muscle activation during the ascending phase of unweighted squats?
- Will the actuator-brace system induce measurable changes in joint kinematics during assisted squatting?
- Does the system maintain natural squat motion without introducing excessive interference or misalignment?

Hypotheses

- **H1:** The hip exoskeleton brace is expected to reduce muscle activation in one or more hip-related muscles during the ascending phase of unweighted squats.
- **H2:** The actuator assistance is hypothesized to produce detectable changes in lower limb joint kinematics, particularly at the hip.
- **H3:** The brace-actuator system is anticipated to preserve natural squat motion without significantly disrupting joint coordination or mobility.

1.4 Significance Of The Study

The outcomes of this research have the potential to contribute significantly to the fields of injury prevention, occupational biomechanics, and assistive device development. By examining how actuator-assisted hip exoskeletons influence muscle activation and joint mechanics during repetitive squatting, this study provides insight into how wearable technologies may reduce biomechanical strain during daily or task-specific movements.

The findings aim to support the development of more ergonomic, lightweight, and accessible exoskeleton solutions for individuals who perform repetitive squatting movements, whether in industrial, clinical, or rehabilitative environments. These insights can inform future guidelines for injury mitigation through the integration of joint-specific assistive technologies.

Moreover, this thesis seeks to bridge the gap between theoretical biomechanical modeling and the practical application of exoskeleton systems. It offers evidence-based data on how hip joint support can affect muscular and kinematic responses during bodyweight tasks, contributing to the evolving field of human-robot interaction in both preventative and supportive movement strategies.



Chapter 2

Literature Review

2.1 Biomechanics Of Bodyweight Squatting And Exoskeletons For Assistance And Injury Prevention

Bodyweight squatting is a fundamental human movement involving coordinated flexion and extension across the hips, knees, and ankles. Biomechanically, a squat requires significant joint range of motion and muscular effort to lower and raise the body's center of mass. During the descent phase (eccentric), the hip and knee joints flex under control of the hip extensors and knee extensors eccentrically, while the ankle dorsiflexes; in the ascent phase (concentric), these joints extend to propel the body upward (Hoogenboom et al., 2023). The movement demands high coordination and stability, engaging multiple muscle groups simultaneously to maintain balance and control. Proper squat technique typically involves keeping the trunk upright (to varying degrees depending on squat style), knees tracking in line with the toes, and feet flat, in order to distribute loads evenly and reduce undue stress on any single joint. A correctly performed bodyweight squat can improve lower-limb strength and neuromuscular control, whereas poor mechanics (such as excessive knee valgus or forward lean) can increase joint stresses and risk of injury.

From a kinetic perspective, squatting generates substantial forces at the knee and hip joints. At maximum depth, compressive forces on the tibiofemoral joint and tension in passive structures (like ligaments) are high, which is why proper form and gradual training progression are emphasized in exercise science (Hoogenboom et al., 2023). Peak knee moments typically occur near the bottom of the squat, requiring strong quadriceps contraction to resist and reverse the motion. Meanwhile, the hip extensors (gluteus maximus and hamstrings) generate large torques to initiate ascent from the flexed hip position. The complexity of the squat's multi-joint biomechanics has been well-documented in literature: for instance, *Lombard's Paradox* describes how biarticular muscles coordinate to allow simultaneous knee and hip extension despite opposing joint actions (this is discussed further in Section 2.2) (Ottinger et al., 2020). In summary, an unassisted bodyweight squat is a demanding compound movement that builds lower-

extremity strength but also imposes high stresses, highlighting the importance of proper technique and, increasingly, the potential role of assistive devices.

Exoskeleton Assistance in Squatting: Wearable lower-limb exoskeletons have emerged as a strategy to assist human squatting movements, either to augment performance or to reduce musculoskeletal strain for injury prevention. Exoskeletons are *wearable robotic technologies* designed to offload forces from the user's body; in industrial and rehabilitation contexts, they can significantly reduce the loads on joints and mitigate the risk of work-related musculoskeletal disorders (WMSDs) (Golabchi et al., 2022). By partially supporting the user's bodyweight or providing external torque at the joints, an exoskeleton can decrease the muscular effort required to perform a squat. For example, a systematic review by Huysamen et al. noted that back-support and leg-assist exoskeletons can lower spinal and knee loading, thereby potentially reducing injury incidence among workers who perform repetitive squatting or lifting tasks (Golabchi et al., 2022). In essence, these devices aim to maintain the natural kinematics of a squat while mechanically assisting movement, thus combining human control with robotic support.

Recent research demonstrates tangible benefits of squat-assist exoskeletons. Metabolic cost and muscle activity tend to decrease when squatting with assistive support (Jeong et al., 2023). For instance, Syed et al. observed that providing body weight support during squats altered biomechanics and reduced muscular demands on the user (though detailed quantitative results depend on support level and device type) (Golabchi et al., 2022) (Li et al., 2025). In one design, a quasi-passive exoskeleton using spring assistance was shown to reduce the user's muscular activity in the quadriceps by roughly 45–72% during ascent, and to reduce plantar pressure under the feet by about 60% (Li et al., 2025). These reductions are substantial, indicating that the exoskeleton bore a significant portion of the load. Consequently, users were able to extend their endurance (holding a squat stance or repeating squats) nearly fivefold – from about 2.8 minutes unassisted to 13.6 minutes with exoskeleton support in that study (Li et al., 2025). Such findings underscore the exoskeleton's effectiveness in reducing fatigue and physical strain.

Moreover, powered exoskeletons can provide personalized assistance profiles during squatting. A study by Chen et al. designed an electromechanical knee-assist system

that actively adapts to the user's squatting motion; it demonstrated that assistance could be synchronized to the squat phases to offload knee extensor muscle effort without impeding natural movement (Li et al., 2025). Another study by Zhang et al. employed a soft pneumatic knee exoskeleton which stored energy in elastomer springs during descent and released it during ascent, yielding reductions in quadriceps muscle activity and knee joint loading (Li et al., 2025). These examples illustrate how assistive strategies (both passive and active) can improve squat biomechanics by reducing peak joint forces and muscle activations, which is directly linked to injury prevention in repetitive tasks. By lowering muscle fatigue and joint stress, exoskeletons help diminish risk factors for injuries such as patellofemoral pain, meniscal strain, or lumbar disc strain that can accumulate over repetitive squatting.

It is important to note that exoskeleton assistance must be carefully controlled to preserve natural movement patterns. Effective assistive squatting devices are typically designed with human biomechanics in mind, aligning with the user's joint axes and allowing free motion when assistance is not needed (Hoogenboom et al., 2023). Misalignment or excessive impedance can lead to abnormal kinematics or discomfort. Therefore, modern squat-assist exoskeletons emphasize ergonomic design – for example, aligning the exoskeleton knee joint with the anatomical knee axis and using compliant or low-impedance actuators (see Section 2.3) to avoid “fighting” the user's own movements (Li et al., 2025). When done properly, the exoskeleton becomes a seamless extension of the body, engaging only to provide helpful torque support or weight offloading, and otherwise allowing the user to maintain control.

In summary, the biomechanics of a bodyweight squat involve complex multi-joint coordination and significant muscular effort. Exoskeletons for squat assistance aim to complement these human biomechanics by reducing loads and augmenting muscle power where needed. Evidence from recent literature indicates that such devices can lower muscle activation levels and joint forces during squatting (Li et al., 2025), thereby prolonging endurance and potentially preventing injuries in occupational or clinical settings. The next sections will explore the specific muscular contributions during squatting (Section 2.2), the technologies enabling exoskeleton assistance (Section 2.3),

and the methods used to analyze muscle activation and movement in such contexts (Sections 2.4 and 2.5).

2.2 Functional Roles Of Key Muscles During Squatting

Squatting recruits a large number of lower-body muscles. This section focuses on five key muscles – Biceps Femoris, Gluteus Maximus, Gluteus Medius, Rectus Femoris, and Tensor Fasciae Latae – describing their functional roles during the squat movement. Each of these muscles contributes to the squat in distinct ways, acting either as prime movers or stabilizers. Understanding their roles is important both for analyzing squat biomechanics and for designing exoskeleton assistance that targets specific muscle groups.

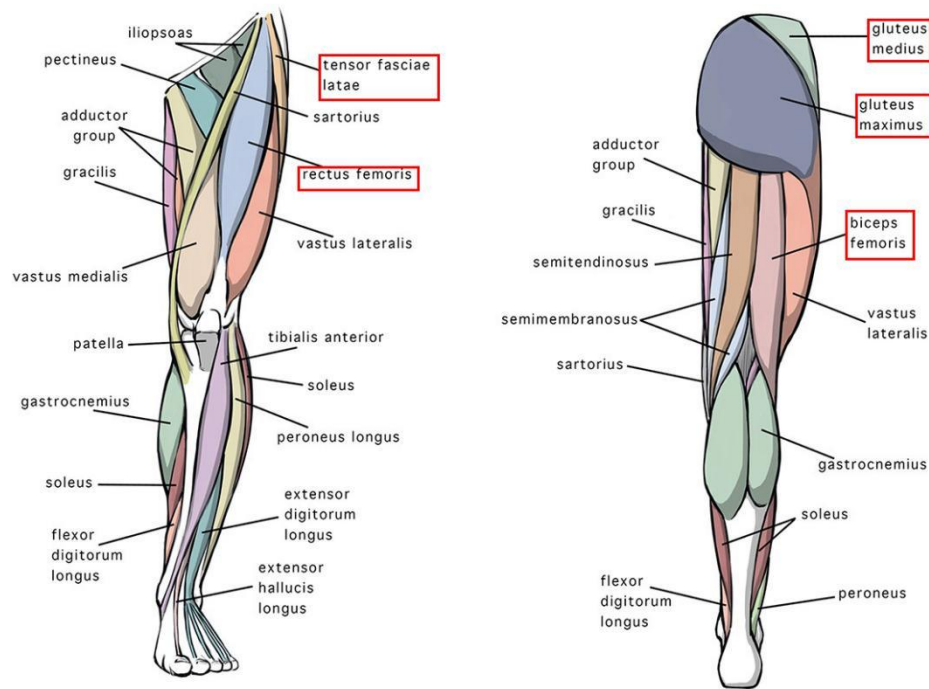


Figure 2. Anterior and posterior views of the leg anatomy (Marzullo, 2022).

2.2.1 Biceps femoris (hamstrings). The biceps femoris, part of the hamstring muscle group, is a biarticular muscle that crosses both the hip and knee joints. Its primary actions are hip extension (long head of biceps femoris, originating on the ischium) and knee flexion (both long and short heads, inserting on the fibula). During the squat, the biceps femoris plays a supportive role in both phases of the movement. In the ascent (upward phase), it assists the gluteus maximus in extending the hip from a flexed position, contributing to raising the torso and hips upward (Physiopedia contributors). This synergistic action helps generate the powerful hip extension needed to stand up from a deep squat. The hamstrings (including biceps femoris) contract to straighten the hip, effectively pulling the pelvis upward and forward in concert with gluteal muscles (Physiopedia contributors).

However, due to its biarticular nature, the biceps femoris experiences Lombard's Paradox during squatting. As the knees and hips move simultaneously, the hamstrings do not undergo dramatic length changes: knee flexion (which would shorten the hamstrings) occurs together with hip flexion (which lengthens them), resulting in a near-isometric behavior over much of the squat's range (Ottinger et al., 2020). Because the muscle length remains relatively constant, the biceps femoris can transmit force effectively without significant shortening, acting almost like a stiff cable linking hip and knee (Ottinger et al., 2020). This allows it to stabilize and coordinate the extension of both joints. Indeed, research suggests that during a squat, the hamstrings are engaged more for joint stabilization and force transmission rather than as prime movers for knee extension (Li et al., 2025). Co-contraction of the biceps femoris with the quadriceps increases knee stability by opposing anterior tibial translation, thereby protecting the knee (and the anterior cruciate ligament) under load (Li et al., 2025).

Notably, EMG studies of barbell squats often find hamstring activation is lower than quadriceps activation, supporting the idea that squats are a quadriceps-dominant exercise and may not maximally recruit hamstrings (Li et al., 2025). The hamstrings' involvement increases with greater forward trunk lean or deeper hip flexion angles, which demand hipper extensor torque. Overall, the biceps femoris contributes to squatting by assisting hip extension and stabilizing the knee, rather than producing large knee flexion

movements during the squat itself. Its quasi-isometric contraction throughout much of the movement helps coordinate the timing of hip and knee extension so that both joints extend smoothly together (Li et al., 2025).

2.2.2 Gluteus maximus. The gluteus maximus is the primary hip extensor and one of the most powerful muscles in the human body. It is critically important in the ascent phase of the squat, where it generates the bulk of the hip extension force. As one lowers into a squat, the gluteus maximus lengthens eccentrically to control the descent; upon rising, it contracts concentrically to drive the hips forward and upward. *Higher squat depths and wider stances* are known to increase gluteus maximus activation (Straub & Powers, 2024). For example, a medium or wide stance squat recruits the gluteus maximus more than a narrow stance, with studies reporting 13%–61% higher gluteus maximus EMG activity in wider stance squats (Straub & Powers, 2024). This is partly due to greater hip external rotation and abduction demands in a wider stance, which engage the glutes to a larger extent. Depth also matters: gluteus maximus activity has been shown to increase substantially from shallow to parallel squats (one study noted a ~65% increase in glute EMG when comparing 40° vs ~90° knee flexion) (Straub & Powers, 2024). At very deep squats, glute activation remains high, although findings vary on whether it plateaus or continues to increase beyond parallel depth (Straub & Powers, 2024). In summary, deep and wide-stance squats elicit the strongest gluteus maximus activation, as these conditions maximize the hip extensor moment requirement.

Functionally, during the upward phase the gluteus maximus works powerfully to extend the hip, helping to lift the torso and hips vertically. It also plays an important role in controlling hip adduction and internal rotation. By pulling the femurs outward (hip abduction/external rotation), the glute max – together with the gluteus medius – helps keep the knees from collapsing inward (valgus) under load (Straub & Powers, 2024). This is crucial for maintaining proper knee alignment and preventing undue stress on the knee joint, particularly on the medial structures. Strengthening the gluteus maximus is often recommended to improve squat form and reduce patellofemoral joint stress, as a strong glute max can counteract knee valgus and internal rotation tendencies (Straub & Powers, 2024). Clinically, trainers often use cues like “spread the floor” or band resistance around

the thighs to encourage greater glute engagement during squats, thereby promoting safer mechanics via enhanced hip external rotation and abduction torque (Straub & Powers, 2024).

In rehabilitative and strength-conditioning literature, the squat is frequently cited as an effective exercise for the gluteus maximus. However, if an individual has dominant quadriceps or weak glutes, they may underutilize the gluteus maximus in favor of other muscle groups (sometimes termed “quadriceps-dominant” squatting). This can manifest as excessive forward knee travel or a lack of hip hinge. Proper coaching (hip hinge technique, sitting back into the squat) encourages gluteus maximus activation. Glute-hamstring synergy is also noteworthy: the gluteus maximus and biceps femoris often work in tandem to extend the hip (Physiopedia contributors). If the gluteus maximus fatigues, research shows the hamstrings can increase their contribution (eccentrically controlling descent more, or aiding hip extension on ascent) (Matsunaga et al., 2021), which is a compensatory strategy but could predispose to hamstring strain if loads are high (Matsunaga et al., 2021). An ideal squat harnesses the gluteus maximus as the prime mover at the hip, which not only produces needed force but also stabilizes the hip and pelvis, contributing to an overall safe movement pattern.

2.2.3 Gluteus medius. The gluteus medius is located on the lateral aspect of the hip and functions primarily as a hip abductor and stabilizer of the pelvis in the frontal plane. During squatting, the gluteus medius is less of a prime mover for the up-and-down motion, and more of a stabilizer that ensures proper alignment of the lower extremities. As one descends into a squat, the gluteus medius on each side works to keep the pelvis level (preventing hip drop or excessive lateral tilt) and to prevent the knees from caving inward. In the ascent, it continues to stabilize and also contributes to hip abduction and external rotation, which helps maintain knee tracking over the toes.

EMG studies indicate that gluteus medius activation is significant during squat exercises, though generally not as high as gluteus maximus activation. The gluteus medius is particularly active when there is a need for frontal plane stabilization – for example, in single-leg squat variations or when there is a side-to-side weight shift. In bilateral squats, gluteus medius activity tends to be moderate. A study by Frigotto et al. (2019) examined

gluteus medius (GMed) vs tensor fasciae latae activation across several lower-body exercises (including single-leg squats) and found that gluteus medius activation was consistently greater than TFL activation in all exercises, including squats (Frigotto et al., 2019). In their results, GMed EMG was classified as *moderate* (on average <40% of maximal voluntary contraction), whereas TFL was *low* (<20% MVC) during bodyweight squat movements (Frigotto et al., 2019). This suggests that the gluteus medius is indeed engaged for stabilization in the squat, taking on more relative load than the TFL. The muscle's role becomes even more pronounced in conditions that challenge frontal plane stability. For instance, during a single-leg squat or an overhead squat, the gluteus medius works hard to prevent the pelvis from dropping and to keep the femurs aligned, which can drive its activation much higher (some studies report gluteus medius activation ranging widely from ~17% up to 80% MVC depending on squat depth and balance demand) (Frigotto et al., 2019).

Functionally, the key role of gluteus medius in squatting is to maintain pelvic and knee stability. Weakness or delayed activation of the gluteus medius often manifests as medial collapse of the knees (valgus) and a hip hike/drop asymmetry. This is a common issue in individuals with knee pain (like patellofemoral pain syndrome) and is a focus in rehab: improving gluteus medius strength has been shown to improve squat and landing mechanics by reducing knee valgus. Thus, the gluteus medius acts to externally rotate and abduct the femur slightly during the squat, counterbalancing internal rotation forces and keeping the knee centered. In the context of exoskeleton assistance or movement analysis, monitoring gluteus medius function is important – an assistive device should not inadvertently destabilize the hip or pelvis. On the contrary, some exoskeletons might even target this muscle by providing lateral support or encouraging proper knee tracking, since the gluteus medius is crucial for safe squat form.

2.2.4 Rectus femoris. The rectus femoris is one of the four quadriceps muscles, unique in that it crosses both the hip and knee joints (originating on the anterior inferior iliac spine and acetabular rim, and inserting via the patellar tendon to the tibial tuberosity). Its primary function is knee extension, with a secondary role in hip flexion. During squatting, the rectus femoris is a key contributor to the knee extensor moment needed to stand up. Along with the other vasti muscles (vastus lateralis, medialis, intermedius), it generates high force to extend the knee from a flexed position. Squatting is well-known to be a quadriceps-dominant action – in fact, the quadriceps are considered the prime movers in a squat, especially through the mid-range and near full extension of the knees (Lee et al., 2022). As one descends, the rectus femoris (and quads) fire eccentrically to control knee flexion; as one ascends, they contract concentrically to straighten the knees.

However, the biarticular nature of rectus femoris introduces interesting mechanics. Similar to the hamstrings, the rectus femoris undergoes counteracting length changes during a squat: as the knee flexes, rectus femoris lengthens at that joint, but simultaneously the hip flexes, which slackens the rectus femoris at the hip. This interplay means the rectus femoris can also maintain a relatively constant length through mid-ranges of the squat – another aspect of Lombard’s paradox, where it transmits force between hip and knee without large changes in fiber length (Ottinger et al., 2020). Consequently, the rectus femoris can act like a *ligamentous tie* between the hip and knee extension motions, coordinating them. It still actively contributes to knee extension torque, but interestingly, research has shown that the rectus femoris may not reach maximal activation during a squat due to this biarticular effect and the fact that other quads (vasti) can provide knee extension force while the rectus assists in hip flexion control. Some studies suggest that at very deep squat positions (hip hyper-flexed), the rectus femoris might be at a mechanical disadvantage (some active insufficiency), causing the vasti to take on more of the knee extension load.

Electromyographic data from various squat techniques indicate the rectus femoris is highly active, though often not the single most activated quad (vastus lateralis and medialis also show high activation). For example, in a bodyweight or light squat, all heads of the quadriceps can reach activation levels on the order of 40–60% MVC or more, with

rectus femoris contributing strongly (Lee et al., 2022). In specialized squat variants, the rectus femoris activation can change: a “Spanish squat” (where one leans against a band or strap at the knee) has been shown to elicit higher rectus femoris EMG than a standard squat, because it shifts emphasis to knee extension over hip extension (Lee et al., 2022). Meanwhile, wall squats or front squats might alter rectus femoris involvement differently. The key point is that rectus femoris is an essential part of the quadriceps action in squatting, extending the knee powerfully. At the same time, through its connection to the hip, it helps maintain hip flexion tension and can stabilize the pelvis relative to the femur.

In summary, the rectus femoris functions as a *biarticular bridge* that aids the quadriceps in knee extension while also contributing to hip flexion control. During ascent, it produces force to straighten the knee; during descent, it resists knee bending and prevents the torso from collapsing too far forward by virtue of its hip attachment. Any exoskeleton assistance aimed at the quadriceps (for example, a powered knee joint) would inherently interact with the rectus femoris’s role, potentially reducing its required output if the device provides knee extensor torque. Care must be taken that such assistance does not inadvertently cause hip instability, given rectus femoris’ dual joint function.

2.2.5 Tensor fasciae latae. The tensor fasciae latae is a smaller muscle on the anterolateral aspect of the hip that inserts into the iliotibial (IT) band. It assists in hip flexion, abduction, and internal rotation, and helps tighten the IT band to stabilize the lateral side of the knee. In squatting, the TFL is generally considered a stabilizer and synergist rather than a prime mover. It contributes to maintaining femur alignment and pelvic stability, working with the gluteus medius and gluteus minimus to balance the hip. When squatting down, the TFL likely engages to counteract external rotation forces and to maintain a balanced hip position (since it internally rotates and abducts the femur slightly). During ascent, it may assist the early phase of hip flexion (when transitioning from the very bottom) by maintaining tension in the IT band, but it is not a major extensor or flexor in this motion.

Studies comparing TFL and gluteus medius activation often find that TFL activation is relatively low in well-performed squats. As mentioned, Frigotto et al. reported TFL activation under 20% MVC in multi-joint lower limb exercises like squats,

classifying it as *low activation*, whereas the gluteus medius was moderate and significantly higher than TFL (Frigotto et al., 2019). This implies that, under normal conditions, the gluteal muscles (maximus and medius) handle the bulk of hip stabilization, and the TFL plays a secondary role. In fact, one concern in rehabilitation is that some individuals might over-recruit TFL in place of the gluteus medius (a pattern seen in certain hip dysfunctions). In squats, however, a healthy movement pattern should emphasize glute medius activity over TFL (Frigotto et al., 2019), which is exactly what the literature demonstrates. The gluteus medius:TFL activation ratio tends to be high, indicating the TFL is not called upon heavily unless the movement or stance specifically biases its function.

Nonetheless, the TFL's functional role in the squat is to provide lateral stability and assist in controlling the hip and knee position. By tensing the IT band, TFL can increase lateral knee stability especially near full extension. At deeper knee flexion, the IT band's influence on knee stability is less direct, but TFL contraction still contributes to hip abduction torque. During a bodyweight squat, if one observes any knee valgus or femoral internal rotation, increased TFL activity might be the body's attempt to counter it (because TFL abducts and internally rotates, it might seem counterintuitive, but co-contraction patterns are complex). Ideally, strong glutes keep the knee aligned and the TFL just offers background support.

In summary, the tensor fasciae latae is minimally to moderately active during squats, mainly serving as a dynamic stabilizer. It works in concert with the larger hip muscles to maintain proper alignment. From an exoskeleton design or training perspective, the TFL is not typically a target for enhancing squat power; instead, it is observed as part of the overall synergy. If an exoskeleton provides lateral support or cues to keep knees out, it might indirectly reduce the need for TFL contribution by enabling or encouraging the gluteals to take the lead. Conversely, any imbalance where an exoskeleton might restrict natural hip motion could potentially cause undue TFL compensation, which should be avoided. Fortunately, most assistive devices focus on sagittal plane assistance (flexion/extension), so the TFL's role remains mostly unchanged – an accessory stabilizer in the background of the squat movement.

Summary of Muscle Roles: In a correctly executed squat, the quadriceps (including the rectus femoris) are the primary knee extensors generating force to lift the body, the gluteus maximus is the primary hip extensor driving the ascent and ensuring an upright trunk, and the hamstrings (biceps femoris) assist hip extension while co-contracting to stabilize the knee joint (Lee et al., 2022). The gluteus medius and TFL act on the side of the hip to maintain lateral stability and proper knee alignment, with the gluteus medius doing the heavy lifting for pelvic stability and the TFL contributing minor support (Frigotto et al., 2019). Understanding these roles provides a foundation for interpreting squat biomechanics data, such as EMG patterns, and for designing exoskeletal assistance that appropriately supports or offloads these muscles. For instance, knowing that the squat is quadriceps-dominant but hamstrings-sparing (Lee et al., 2022) helps explain why certain exoskeletons focus on knee extension assistance, and recognizing the gluteus medius's importance in stability might lead to designs that do not impede lateral hip movement. In the following sections, we delve into the technologies for exoskeleton actuation (Section 2.3) and the methods (EMG and motion capture) used to analyze the complex interplay of these muscles and joints during squatting (Sections 2.4 and 2.5).

2.3 Actuator Technologies In Exoskeletons And The Role Of Gear Ratio (9:1) In Performance And Safety

Exoskeletons rely on actuators to generate forces or torques to assist human movement. The choice of actuator technology is a pivotal design decision, affecting an exoskeleton's power output, responsiveness, weight, noise, and safety. Broadly, the actuators used in exoskeletons can be classified into three categories: electric, pneumatic, and hydraulic (active actuation), with some devices also employing passive elements like springs or dampers (passive actuation) (Urrea & Agramonte, 2023). Each technology has distinct advantages and trade-offs. This section provides a comparative review of electric vs. pneumatic vs. hydraulic actuators in exoskeleton applications, and examines how the gear ratio – particularly a 9:1 transmission – influences output torque, control responsiveness, and user safety in electric motor-driven systems.

2.3.1 Electric actuators (motors and gear drives). Electric actuators (typically DC or brushless motors) are the most common choice for wearable exoskeletons, owing to their precision, relative compactness, and ease of control (Urrea & Agramonte, 2023). Electric motors provide rotary motion and, when coupled with a transmission (gearbox or pulley system), can produce the high torques needed to assist human joints. They also interface well with electronic control systems for finely tunable assistance profiles. A key design parameter in electric actuation is the gear ratio of the transmission between the motor and the exoskeleton joint.

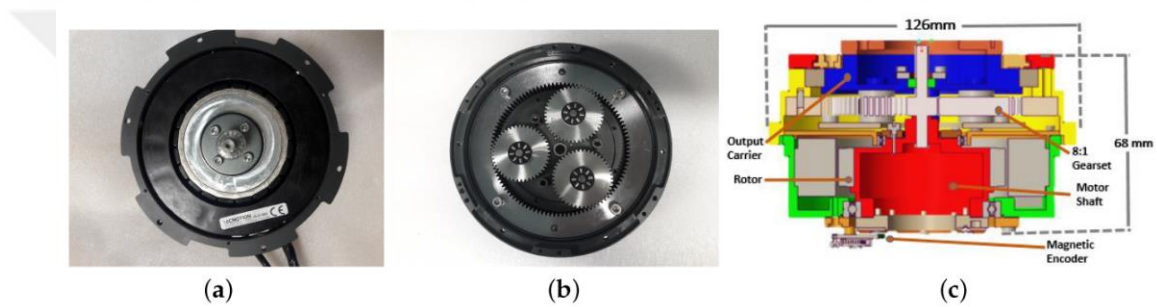


Figure 3. From left to right — (a) motor sub-assembly, (b) gearbox sub-assembly, and (c) cross-sectional cad view of the complete actuator assembly, designed to fit within dimensions of 126×68 mm (Singh et al., 2020).

High gear ratios (e.g., 50:1 or 100:1) allow a small, high-speed motor to produce large output torques by trading off speed. However, high ratios also amplify the motor’s reflected inertia and friction at the output, making the joint stiffer and less backdrivable (harder for the human to move freely when the motor is unpowered or when resisting the motor) (Zhu et al., 2022). This can hinder natural movement and pose safety concerns – if the exoskeleton applies force in an undesired way, a high ratio transmission might resist the user’s attempt to correct or escape that motion. High ratios also often introduce backlash (small play in gears) and can reduce the responsiveness of the system to rapid changes, due to motor speed limitations and compliance issues.

Low gear ratios (e.g., 5:1, 9:1, or “quasi-direct-drive” transmissions $\leq 10:1$) have become a favored approach in recent exoskeleton designs to improve transparency and safety (Zhu et al., 2022). A low gear ratio means the motor must produce more torque directly (often requiring a larger or more powerful motor), but the benefit is dramatically

lower reflected inertia and friction at the joint. This yields a highly backdrivable and compliant actuator – one that the user can move with relatively low effort when needed, and that can quickly respond to and accommodate unexpected disturbances (Zhu et al., 2022). In other words, low ratios improve the intrinsic compliance of the actuation, enabling the exoskeleton to “get out of the way” of the user’s natural movements and reducing the risk of the device overpowering or misaligning the user.

The concept of Quasi-Direct Drive (QDD) refers to using a low gear ratio with a high-torque motor to achieve a balance of torque amplification and compliance (Zhu et al., 2022). A gear ratio around 9:1 is a canonical example of QDD in current exoskeleton research. For instance, one hip exoskeleton project selected a commercial actuator (the AK80-9) which has a built-in 9:1 planetary gearbox, citing its combination of high torque density and low gear ratio (9:1) as ideal for wearable use (Nesler et al., 2022). This actuator (80 mm motor, 9:1 reduction) can deliver about 9 Nm continuous and 18–22 Nm peak torque in a compact 0.5 kg package. The low ratio keeps output impedance low, meaning the exoskeleton joint can be back-driven with minimal resistance, enhancing safety. One study noted that this actuator’s specs – *9:1 ratio, ~485 g mass, ~233 cm³ volume* – made it well-suited for hip exoskeletons because it provided sufficient torque at low inertia and even relatively low cost (Nesler et al., 2022).

Effect of a 9:1 Gear Ratio on Torque and Responsiveness: A 9:1 gear ratio means the motor will rotate 9 times for each one rotation of the output joint. Torque is roughly amplified by a factor of 9 (minus losses), while reflected inertia is amplified by the square of the gear ratio (81 times motor inertia reflected). For a well-chosen motor, 9:1 is low enough that the reflected inertia is still small in absolute terms. Research by Zhang et al. on a knee prosthesis with QDD actuation demonstrated that using a ~9:1 total gear reduction (in their case, a 9:1 internal + additional 4:1 stage for 36:1 total) allowed the system to achieve 200 Nm peak torque while keeping backdrive (residual) torque as low as 2.6 Nm (Nesler et al., 2022). In practice, 2.6 Nm backdrive torque is extremely low relative to 200 Nm output – this indicates very high compliance. The authors highlight that motor inertia and gear ratio are the key factors in minimizing backdrive torque, and that (Zhu et al., 2022). In essence, a 9:1 ratio hits a sweet spot: it gives a moderate

mechanical advantage to boost torque, but not so much as to severely compromise the system's ability to yield to external forces.

With a 9:1 (or similar low) ratio, the responsiveness of the exoskeleton also tends to improve. The actuator can accelerate the joint faster because there is less gear reduction limiting the output speed. Also, any control command results in more immediate motion at the joint (since only a 9:1 difference vs motor, instead of, say, 100:1). Combined with modern control algorithms, such systems can achieve high-bandwidth torque control. For example, quasi-direct drive actuators can be controlled to track desired torques or positions very rapidly, often enabling feedback loops that account for the user's intention (detected via sensors) almost instantaneously. This responsiveness is critical for user safety – the exoskeleton can quickly adjust or shut off torque if it detects a mismatch with user movement, and the low mechanical impedance ensures that even in the absence of motor power, the user can move the joint reasonably freely.

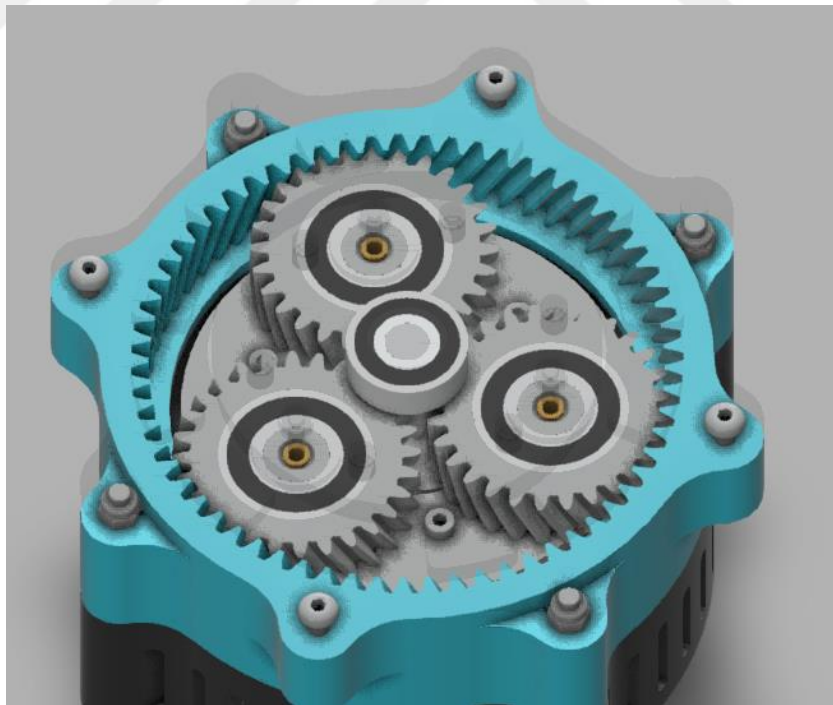


Figure 4. Illustrates a planetary gear system (also known as an epicyclic gear train), which consists of a central sun gear, multiple planet gears, a surrounding ring gear with internal teeth, and a planet carrier that holds and supports the planet gears.

From a safety standpoint, the low gear ratio paradigm significantly reduces risks in human-robot interaction. If an exoskeleton joint with a 9:1 ratio encounters an unexpected obstruction or if the user moves contrary to the motor's action, the resulting forces transmitted to the user are much smaller than in a high-ratio system (Zhu et al., 2022) (Urrea & Agramonte, 2023). The device is more *forgiving*: it will backdrive (i.e., allow the user to win) with relatively low resistance, which is a form of intrinsic safety. High gear ratio systems, by contrast, can lock up or impart large forces before yielding, which could lead to joint injury or falls in worst cases. For these reasons, many state-of-the-art exoskeletons opt for either low-ratio direct drives or series elastic actuators (SEAs), both of which prioritize safety and compliance. SEAs, in particular, incorporate an elastic element to absorb shocks and limit force, achieving a similar goal of protecting the user (Urrea & Agramonte, 2023). But SEAs add complexity; QDD with $\sim 9:1$ gear ratio offers a simpler mechanical solution to get low output impedance (Zhu et al., 2022).

It's worth noting that a gear ratio of 9:1 still requires careful motor selection. The motor must have high torque output (and typically high torque *density* to keep weight low). Advances in motor design (e.g., outrunner brushless motors with large diameters) have made it feasible to have strong motors that don't rely on large gear reductions (Nesler et al., 2022). For example, motors with torque constants allowing $\sim 10\text{--}20$ Nm directly can, with a 9x gearbox, output 90–180 Nm – suitable for knee or hip assistance. These quasi-direct drives often use high-pole-count brushless motors and low-friction gearboxes (like harmonic drives or planetary gears with high efficiency). They also produce less noise and have fewer points of failure than a complex multi-stage high ratio gearbox.

In summary, electric actuation with a $\sim 9:1$ gear ratio provides a compelling mix of high torque, fast responsiveness, and safety through compliance. This approach has been demonstrated in research prototypes achieving human-scale torques with low effective impedance (Li et al., 2025) (Zhu et al., 2022). Electric actuators are also relatively easy to control in a closed-loop, allowing sophisticated algorithms (e.g., impedance control, hybrid position/force control) that further enhance safety by actively modulating how the exoskeleton interacts with the user (Li et al., 2025)(Zhu et al., 2022). The result is an

actuator technology that aligns well with the needs of assistive squatting: strong yet yielding support that can be precisely timed with the user's effort.

2.3.2 Pneumatic actuators (artificial muscles and air cylinders). Pneumatic actuators use compressed air to produce force and movement. In exoskeletons, two common forms are pneumatic cylinders (air pistons) and pneumatic artificial muscles (PAMs), such as McKibben muscles (Zhong et al., 2020). These systems have the advantage of inherent compliance – air is compressible, and pneumatic muscles are deformable – which makes them shock-absorbing and safer in human interaction by nature. They also boast high power-to-weight ratios; for example, PAMs can generate large forces relative to their mass and contract in a muscle-like manner (Urrea & Agramonte, 2023). This biomimetic property has drawn interest for exoskeletons aiming to replicate the smooth, natural force application of human muscles (Urrea & Agramonte, 2023).

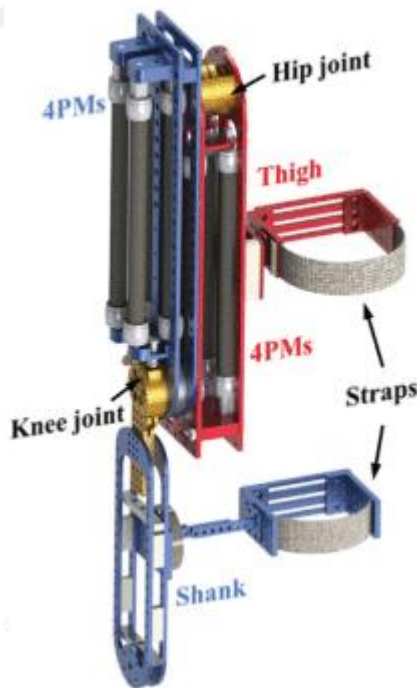


Figure 5. Pneumatic artificial muscles exoskeletal design (Zhong et al., 2020).

Key features of pneumatic actuators:

Compliance and Safety: The compressibility of air and elasticity of pneumatic muscles mean that these actuators are effectively spring-like. They do not have a fixed gear train, so if a user encounters resistance or the actuator overshoots, the air can compress or the bladder can stretch, buffering the interaction forces. This yields a kind of low mechanical impedance that is similar to (or even greater than) what electric SEAs achieve (Urrea & Agramonte, 2023). From a safety perspective, this compliance is valuable because it reduces the risk of sudden hard impacts or excessive torques on the user's joints.

Power and Speed: Pneumatics can be very powerful – industrial pneumatic cylinders can lift hundreds of kilograms – and can also respond quickly, limited mainly by valve switching speed and air flow rates. PAMs, for instance, have been shown to produce high force with a fast contraction speed, making them capable of (Urrea & Agramonte, 2023). They operate within typical pressure ranges (e.g., 0–8 bar) and can contract by shortening when pressurized, similar to a muscle's pull.

Weight and Portability: A challenge with pneumatic systems in mobile exoskeletons is the need for an air supply (compressor or pressurized tanks). While the actuators themselves (cylinders or muscles) are light, the ancillary equipment (pumps, valves, hoses) can add weight and reduce portability. Some exoskeleton designs use off-board air supplies (tethered setups) for lab studies, which is fine for research but not for real-world use. Others incorporate lightweight compressors or use CO₂ cartridges, but these have limited operation time. In industrial settings with stationary air sources, pneumatics can be viable.

Control and Precision: Pneumatic actuators are non-linear and have hysteresis, especially PAMs (Urrea & Agramonte, 2023). The force output of a PAM depends on the current pressure and its contraction (length) state, making the control more complex than electric motors (which have linear torque constant regions). Precise position or torque control with pneumatics is challenging; techniques include advanced model-based control, sliding-mode or fuzzy controllers, as well as adding feedback sensors to reduce

uncertainty (Urrea & Agramonte, 2023). Achieving high accuracy in synchronization with human motion (like squatting) requires careful tuning. Nonetheless, researchers have been improving these aspects, and pneumatics remain attractive for their muscle-like feel.

In the context of squat-assist exoskeletons, pneumatic systems have been explored in both passive and active forms. Some passive exos rely on pneumatic springs (gas springs) that provide a constant upward force, effectively partially counteracting body weight in a squat. Active pneumatic systems might use a control valve to modulate air into a cylinder at the knee joint: e.g., inflate to help stand up, vent to allow bending down. A noteworthy example is the use of McKibben pneumatic muscles (Zhong et al., 2020) at the hip or knee; when the user squats, the muscles stretch and then upon command they are pressurized to contract and help lift the user. Pneumatic muscle actuators can closely mimic the tension-length curve of human muscles, providing a very natural assistance profile when do correctly (Urrea & Agramonte, 2023).

Comparatively, pneumatic vs. electric: Pneumatic actuators, without a rigid transmission, don't have a "gear ratio" in the same sense. The concept of gear ratio's effect (like the 9:1 discussed above) is mostly relevant to geared electric drives. Pneumatics instead have an effective stiffness and force-length behavior. But we can analogously note that a PAM's braid angle and geometry sort of determine its force amplification versus contraction (you could call that an intrinsic "gear" of sorts). Generally, a single PAM can create high force without mechanical gearing, so long as the air pressure is available. The responsiveness of a pneumatic exo comes down to valve speed and air dynamics; modern servo valves can change pressure quickly, but there is always a slight lag as air flows and pressure equilibrates. This makes ultra-fast control (millisecond-scale) more difficult, and the system can feel a bit "springy" or delayed compared to an electric motor that is directly driven.

From a safety perspective, pneumatic systems inherently avoid large gear ratios and their issues; they are among the safest in terms of not resisting the user unexpectedly (because if a user moves faster or differently, air can move in/out or the muscle can slacken). However, one safety concern is that if a high-pressure surge occurs (faulty valve or control), a pneumatic could in theory jolt – but most exoskeleton designs include

pressure regulators and consider fail-safes (like valves that open on power loss to release pressure).

In summary, pneumatic actuators in exoskeletons offer excellent compliance and biomimetic force but at the cost of more complex infrastructure and control. They shine in applications where soft, smooth interaction is paramount. Indeed, some researchers envision hybrid systems: electric motors for precise control at some joints and pneumatic muscles for powerful, soft assistance at others (Urrea & Agramonte, 2023). For squatting assistance, a pneumatic exoskeleton might allow a very comfortable user experience – feeling like someone is gently lifting you – but designers must manage the bulk of air components and ensure control fidelity.

2.3.3 Hydraulic actuators. Hydraulic actuators use pressurized fluid (typically oil) to generate motion, often through cylinders (pistons) or hydraulic motors. Hydraulics are known for their high force and power density – they can produce very large forces in a relatively small package and are common in heavy machinery. In the exoskeleton realm, hydraulics has seen use primarily in high-load or military exoskeletons and some rehabilitation devices. For example, early powered exoskeleton prototypes like the BLEEX (Berkeley Lower Extremity Exoskeleton) (Sun et al., 2021) and the Sarcos Guardian XO featured hydraulic actuation to enable lifting of heavy loads.

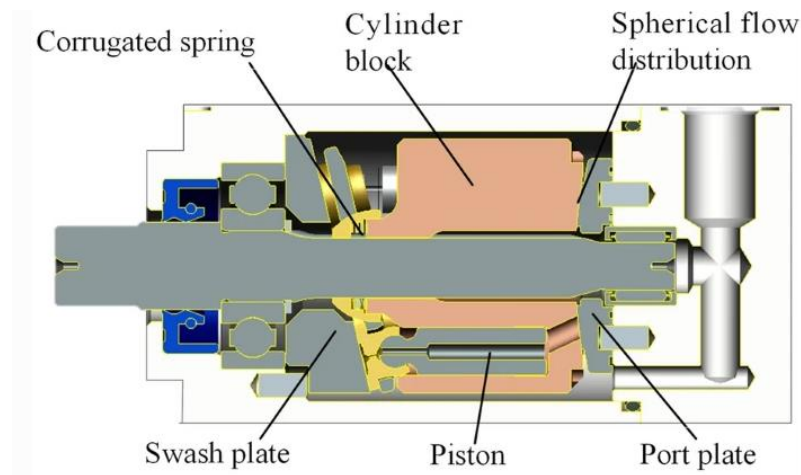


Figure 6. Cross-sectional view of a swash plate axial piston pump. Exemplifies how hydraulic systems can efficiently convert fluid power into linear or rotary motion in compact actuator designs (Sun et al., 2021).

Key characteristics of hydraulic actuators for exoskeletons include:

Superior force output: Hydraulics can easily achieve the torque and force requirements of human joints (hundreds of Nm) with compact actuators. A small hydraulic cylinder can lift a person's weight without strain, which is advantageous for heavy-duty assistive exoskeletons (e.g., those intended to help a worker carry extreme weights or a soldier augmenting strength).

Stiffness and Control: Hydraulic systems tend to be very stiff (incompressible fluid, little give), which on one hand means precise position control is possible, but on the other hand, this lack of compliance can be a safety issue for human interaction. Modern approaches incorporate servo-valves and accumulators to modulate this stiffness and introduce some compliance if needed, but fundamentally, hydraulics are less backdrivable – if the valve is closed, the fluid locks in place and the actuator will resist motion. This is great for holding a static load with minimal energy (just close a valve and the exo could lock a squatting person in position), but not great if the person wants to move quickly or if something goes wrong.

Size and Complexity: The downsides of hydraulics in wearable systems are their bulk, weight, and complexity (Urrea & Agramonte, 2023). A hydraulic exoskeleton requires pumps (or a high-pressure supply), fluid reservoirs, valves, and hoses. These components add significant weight and noise (pumps can be loud). They also require maintenance (avoiding leaks, changing fluid) and can be expensive. For a squat-assist exoskeleton which ideally should be lightweight and ergonomic, hydraulics are usually considered impractical unless absolutely necessary for the force requirements.

Responsiveness: Hydraulics can respond very fast in terms of fluid dynamics (used in fast machinery), but in a wearable, the control is usually slower due to the need to ensure stability (sudden valve changes can cause jerky motions). There is also some lag due to fluid inertia and compressibility of small amounts of entrapped air, etc. Thus, controlling a hydraulic exo to gently mirror human movement is a real challenge – advanced feedback control (like using model-based or adaptive controllers) is needed to tame the system (Urrea & Agramonte, 2023).

From a safety standpoint, hydraulics poses unique challenges. Because they can exert very large forces, any control error can be dangerous. If a valve sticks or a sensor fails, a hydraulic actuator might unintentionally apply a huge force. Therefore, researchers have developed robust control strategies and fail-safes for hydraulic exoskeletons. For example, Yang et al. implemented a disturbance observer-based sliding mode controller on a hydraulic knee exoskeleton to handle uncertainties and prevent unstable force outputs (Urrea & Agramonte, 2023). Similarly, fault-tolerant control systems have been proposed to immediately release pressure or default to a safe state if irregularities are detected (Urrea & Agramonte, 2023). Additionally, many hydraulic exoskeletons incorporate pressure relief valves that limit the maximum force.

In practice, wearable hydraulic exoskeletons are less common nowadays in academic literature, except in specific high-force use cases, due to the rise of powerful electric motors and the difficulties mentioned. One recent trend is the development of compact electro-hydrostatic actuators (EHA) – self-contained units combining motor, pump, and cylinder – which try to bring hydraulic-level forces in a modular form. But those are still heavy for something like a knee assist device. Passive hydraulics (like damping cylinders) are used in some exos for energy absorption or smooth motion (e.g., a damper to control squat speed), which is a simpler use of fluid power.

Comparatively, where an electric system might use a 9:1 gearbox to get torque, a hydraulic system might achieve the same or more torque with a cylinder and no mechanical reduction, just fluid pressure. The concept of gear ratio in hydraulics would relate to cylinder piston area ratios or motor displacement – effectively, a larger piston area gives more force (like a higher gear), but moves a shorter distance for a given fluid volume (like slower speed). Designers pick cylinder sizes to match the needed force and stroke (range of motion). A too-large cylinder gives great force but needs lots of fluid to move (slower, more pump work), whereas a small cylinder moves fast but with less force. This trade-off is analogous to gear ratio choices in motors.

To illustrate, a hydraulic squat-assist could use a cylinder at the knee such that when the user squats, fluid is pushed out to an accumulator, and when the user stands, the fluid is pushed back in to help. If the cylinder is sized to output, say, 1000 N force at the

knee (which might correspond to supporting a significant portion of body weight), the system will be tuned so that it provides that assistive force with a comfortable motion profile. Ensuring the user safety would involve careful limits on pressure (so it never extends too quickly or too far) and giving the user override ability (e.g., manual release or enough transparency so they can still move if needed).

In summary, (Urrea & Agramonte, 2023). They excel in heavy-load contexts (like powered exoskeletons intended to amplify human strength), but for movement assistance in daily tasks, their practicality is limited. The trend in the field has been toward electric and soft actuators, except for specialized cases. For completeness, it's noted that new forms of actuators (like electrohydraulic hybrids or even smart materials like shape memory alloys) are being researched for exoskeletons (Urrea & Agramonte, 2023), but these are beyond the scope of the current discussion.

2.3.4 Gear ratio considerations across technologies. While gear ratio is a term most directly applicable to electric motor gearboxes, the underlying consideration – the balance between torque amplification and system impedance – is universal. In electric systems, a 9:1 gear ratio was highlighted as a design choice to ensure high torque with low impedance (Zhu et al., 2022). In pneumatic systems, one might consider an analogous “gear” as the leverage of a linkage or the diameter of a pneumatic cylinder (larger diameter = more force, less stroke speed – akin to a higher gear ratio yielding more force, less speed). In hydraulic systems, the piston area or hydraulic motor displacement plays a similar role.

The common theme is that increasing force output via mechanical advantage typically comes at the cost of reduced responsiveness or back drivability, so a balance must be struck. All exoskeleton designers must consider this trade-off:

For electric actuators: choose a gear ratio (or use direct drive) that provides enough torque but not so high as to feel rigid or unsafe. The 9:1 example shows a near-optimal point for many lower-limb joints (Zhu et al., 2022) (Li et al., 2025).

For pneumatic actuators: ensure the muscle or cylinder can deliver required force without being so large as to slow the system or make it hard to control. If a PAM is too strong (at high pressures) and not suitably regulated, it could overshoot – hence control valves act like the “gearing” to throttle force.

For hydraulic actuators: use the smallest piston that meets force needs to avoid unnecessarily high impedance. Also incorporate accumulators or valves to allow free motion when full power isn’t needed, effectively making the system “back drivable” in certain modes (some advanced exos have done this by using open-center valves for transparency).

Summary of Actuator Comparison: Electric motor-driven actuators (with appropriate gearing, e.g., 9:1) are currently the most prevalent in squat-assist exoskeletons due to their precise controllability and improving power densities. Pneumatics offer a compelling compliant alternative that mimics muscle behavior, though with added bulk and control complexity. Hydraulics provide brute force for extreme applications but are generally less favored for everyday wearable devices due to weight and safety management issues (Urrea & Agramonte, 2023). Table 2.1 (hypothetical) could summarize these points, but in prose: an active exoskeleton can use *electric, pneumatic, or hydraulic actuators*, and each choice must ensure user safety by balancing output and impedance (Urrea & Agramonte, 2023). Modern trends even combine approaches, such as electric motors driving PAMs or series elastic elements to get the best of both worlds (Urrea & Agramonte, 2023).

Ultimately, regardless of actuation type, user safety and comfort are paramount. A recurring design principle is to incorporate compliance either mechanically (low gear ratio, elastic elements, compressible fluid) or via control (force/torque control, impedance control) so that the exoskeleton doesn’t harm the user (Zhu et al., 2022). The focus on a 9:1 gear ratio in literature highlights how even in high-performance systems we deliberately limit mechanical advantage to keep the system inherently safe and backdrivable. As exoskeleton technology advances, we expect actuators to become lighter, more “intelligent,” and better tailored to human biomechanics – for example, future squat-assist actuators are anticipated to be multi-modal, lightweight, and customizable to

specific tasks (Li et al., 2025). This could mean hybrid electric-pneumatic devices or new materials, but the core challenge remains: delivering just the right assistive force at the right time, while preserving the user's natural control and safety.

2.4 Surface EMG Signal Analysis In Human Movement Studies

Surface electromyography (sEMG) is a fundamental tool for assessing muscle activation in biomechanics research, including studies of squatting and exoskeleton assistance. By placing electrodes on the skin over a muscle, researchers can record the electrical signals associated with muscle fiber depolarization. The raw EMG signal is an oscillating voltage (in millivolts) reflecting motor unit action potentials. However, raw EMG is often too noisy or complex to interpret directly, so it undergoes processing and feature extraction to quantify aspects of muscle activity. This section details several common EMG time-domain features – including Root Mean Square (RMS), Mean Absolute Value (MAV), Area Under the Curve (AUC), Variance (VAR) – and explains their relevance in human movement studies.

EMG signal features can characterize the intensity, duration, and frequency content of muscle activation. They are widely used both in clinical gait/squat analysis and in engineering applications like pattern recognition for prosthetic control (Ahsan et al., 2011). Below is an overview of each feature:

Root Mean Square (RMS): RMS is essentially the standard deviation of the EMG signal amplitude, providing an estimate of the signal's power. It is calculated as the square root of the mean of the squared amplitudes over a window of time. RMS is one of the most representative measures of EMG amplitude and correlates with muscle contraction force in isometric conditions (Ahsan et al., 2011). A higher RMS value indicates a higher muscle activation level (more motor units firing and/or firing at higher rates). Because squatting involves varying muscle force throughout the movement (e.g., higher at the bottom of the squat), RMS can be computed over specific phases to compare effort. In dynamic tasks like squatting, RMS is often used as an average measure of muscle activity magnitude during the ascent or descent phase (Van den Tillaar, 2019).

Mean Absolute Value (MAV): MAV is the mean of the absolute values of the EMG signal over a window (Zhou et al., 2019). Effectively, this is the average rectified EMG amplitude. MAV is closely related to RMS (for a zero-mean signal, $RMS = \sqrt{\text{mean of squares}}$ vs $MAV = \text{mean of absolute}$; they differ slightly in how they weight larger amplitudes). MAV is simple to compute and, like RMS, reflects the overall level of muscle activation. It is often used in real-time applications because it's computationally less intensive (no squares or square-roots) and provides a smooth envelope of the EMG. In squat analysis, MAV can indicate how active a muscle was on average during, say, the lowering phase versus the lifting phase.

Area Under the Curve (AUC): In EMG terms, AUC usually refers to the integrated EMG (IEMG), which is the time integral of the rectified EMG over a period. Essentially, it's the sum of absolute EMG values over time, equivalent to the area under the EMG amplitude-time curve (Zhou et al., 2019). AUC represents the total activation over a movement or trial. For instance, if one performs a squat and we integrate the EMG from start to finish, the AUC gives a measure of the total muscular work (neuromuscular effort) done by that muscle in that rep. It is useful for comparing fatigue or workload: a higher AUC might mean a muscle did more total work. In rehabilitation or exoskeleton studies, one might compare AUC of EMG for a muscle with and without assistance to see if the exoskeleton reduced the muscle's workload over the motion.

Variance (VAR): This is the statistical variance of the EMG signal amplitude in a given window (Hamed et al., 2012). Because EMG is zero-mean (after removal of DC offset), variance is directly related to the power of the signal (power = variance for zero-mean signal). Thus, variance is another way to quantify EMG intensity. It gives more weight to larger deviations (squared term) and is used in some pattern recognition algorithms as a feature. In a practical sense, higher variance means more fluctuation in the EMG – likely corresponding to stronger or more phasic muscle contractions. For instance, an erratic firing or a very strong contraction burst will increase variance. In contrast, a steady low-level contraction yields low variance. When analyzing squat biomechanics, researchers often calculate many of these features for each muscle's EMG to quantify the muscle's behavior. For instance, one might report that during the ascent phase of a squat,

the RMS EMG of the rectus femoris was 0.15 mV (indicating high activation), the IEMG (AUC) over the whole squat was X (indicating total effort), and maybe compare SSC or mobility to assess if the muscle activation was more burst-like or steady. In exoskeleton studies, EMG features are crucial for evaluating assistance – a successful exoskeleton might reduce RMS and AUC of the user’s EMG for a given task, meaning the muscle worked less hard overall (Li et al., 2025). On the other hand, features like complexity or SSC might inform whether the assistance caused any abnormal activation patterns (one would hope not).

Additionally, EMG features are often inputs to machine learning models or control systems. For example, a controller might use MAV or RMS in real-time to gauge the user’s effort and adjust assistance accordingly. Pattern recognition systems for prosthetic or exoskeleton control frequently use a set of time-domain features (MAV) as the input vectors to classify movements (Ahsan et al., 2011). In recent literature, there’s also interest in using more advanced features or deep learning on raw EMG, but the classical features remain very interpretable and useful.

In summary, surface EMG signal analysis in movement studies relies on extracting meaningful features that condense the raw EMG into quantifiable metrics of muscle activity. Features like RMS and MAV capture amplitude (related to force output), AUC captures total effort, VAR capture signal power and complexity (Zhou et al., 2019). Together, these paint a comprehensive picture of muscle function. For the squat and exoskeleton context, these features allow researchers to quantify how an exoskeleton affects muscle activation patterns – for instance, confirming that assistance reduces muscle RMS (less effort) without causing abnormal spikes or co-contraction (which might show up in features like SSC or complexity). Proper use of EMG analysis thus supports both the efficacy and safety evaluation of assistive devices.

2.5 Motion Capture In Squat Biomechanics And Kinematic Feature Analysis

Motion capture technology is a cornerstone of biomechanical analysis, providing precise measurements of human movement kinematics. In squat biomechanics research, 3D motion capture (using optical marker-based systems or newer inertial sensor systems)

allows for detailed tracking of joint angles, segment orientations, and movement trajectories throughout the squatting motion (Hoogenboom et al., 2023). This section discusses the application of motion capture in analyzing squats and how kinematic features such as Range of Motion (ROM), and analogues to EMG features like RMS and MAV of joint angle/velocity, are used for joint analysis.

Using motion capture, researchers can reconstruct the movement of each major joint during a squat. Key kinematic variables of interest often include the angles at the hip, knee, and ankle over time, as well as trunk inclination. These angle-time plots characterize the squat technique – for example, a deeper squat is reflected by larger knee and hip flexion angles. MoCap data can also be combined with force plate data to compute joint kinetics (moments and forces) via inverse dynamics, although here we focus on the kinematics.

Range of Motion (ROM): One fundamental measure is the ROM of a joint during the squat, defined as the difference between the maximum and minimum angle attained. For instance, if the knee goes from nearly full extension (0° flexion) at the top to 90° flexion at the bottom, the knee ROM is $\sim 90^\circ$. ROM is an indicator of squat depth and flexibility. In literature, greater ROM (especially at the knee and hip) is often associated with more muscle recruitment and different loading patterns (Straub & Powers, 2024). Squat depth is sometimes categorized (partial, parallel, full/deep) based on knee or hip ROM achieved. Motion capture provides accurate ROM measurements, which are used to standardize comparisons (e.g., comparing biomechanics at 90° knee flexion vs 120° knee flexion squats) (Straub & Powers, 2024). ROM is also clinically relevant: insufficient ankle ROM (due to tight calves) might cause heel rise or forward lean; limited hip ROM might prevent deep squatting. Thus, MoCap helps identify such restrictions by quantifying each joint's motion envelope. In exoskeleton studies, one ensures that the device does not limit the user's natural ROM – motion capture can verify that the user still achieves a normal ROM with the exo on.

Time-Domain Features of Kinematic Signals: Just as we extract features from EMG signals, similar analyses can be applied to kinematic data (joint angle vs time, or joint angular velocity vs time). While less common than EMG features, these can provide

insights into the character of the movement beyond simple discrete points like ROM or peak velocity. For example:

The RMS of a joint angle signal during a movement could be interpreted as a measure of the typical deviation of that joint from its neutral position. However, because joint angles are not zero-mean signals (e.g., an angle might oscillate around some offset), RMS is less straightforward. It's more meaningful to apply RMS to velocity or acceleration signals, which do oscillate around zero when the movement is back and forth. For instance, one might calculate the RMS of the knee angular velocity during the descent phase. A higher RMS velocity would indicate a faster, more dynamic descent on average, whereas a low RMS velocity indicates a controlled, slow descent. RMS angular velocity might correlate with things like momentum and impact forces (fast squats might lead to higher impact at bottom).

MAV (Mean Absolute Value) of joint velocity would be analogous to average speed of movement. For example, if one squats at an average speed of $30^{\circ}/s$ versus $60^{\circ}/s$, the MAV of angular velocity would reflect that difference. This could be relevant when comparing controlled vs ballistic squats, or evaluating if an exoskeleton encourages a smoother constant-speed lift versus a more erratic motion.

Range of Motion (ROM) we discussed, but also Peak angles and Peak angular velocities are important discrete kinematic outcomes. Peak angular velocity of the knee during ascent might relate to how explosively the person stood up (useful in athletic performance analysis). Motion capture data often yield such values which are compared across conditions. For example, How Are Squat Timing and Kinematics related to depth? A study might find deeper squats have slower average ascent velocity, or a longer descent phase (Zawadka et al., 2020).

In some advanced analyses, researchers use principal component analysis (PCA) or other methods on joint angle time-series to capture movement patterns. But focusing on simpler features: one could treat the angle-time curve like any signal and compute e.g. the variance of the joint angle over the trial (which partly is determined by ROM), or the

integral of absolute angular velocity (which would equal total angular distance traveled, effectively related to ROM as well).

For squat biomechanics, an important application of motion capture is assessing movement quality and symmetry. Kinematic features can reveal compensations: for instance, if an exoskeleton is too restrictive at the hip, one might see abnormal increases in lumbar flexion or something – MoCap would catch that by comparing ROM at those joints with and without the device. Researchers also look at coordination timing (e.g., the timing between hip and knee extension, sometimes called inter-joint coordination). If a device altered how a person sequences their joint extension, it could show up in phase plots or relative angle analyses. This might be beyond simple RMS/MAV, but motion capture data can be used to derive such coordination metrics.

Another set of kinematic features used especially in screening or assessing squat performance are things like trunk lean angle, knee travel, and pelvic tilt. These are taken at key instants (like at the deepest point). They help evaluate technique. For example, excessive forward trunk lean (angle of torso from vertical) can indicate poor ankle flexibility or weak quads forcing a hip-dominant strategy. Motion capture studies often report these angles (Straub & Powers, 2024). In an exoskeleton context, one might ensure that with assistance, users do not develop altered strategies like leaning more (which could offload work to the device but strain the back).

Use of Kinematic RMS/MAV in Research: While not as commonly reported as EMG features, some research does incorporate measures like mean or RMS joint speeds. For instance, a study might report the mean descent velocity of the squat, which is effectively MAV of the angular velocity across the descent phase (Brice et al., 2022). Similarly, if analyzing variability, one could compute the standard deviation of the joint angle over multiple squat trials to assess consistency. In rehabilitation, a smooth control of movement might be reflected in a lower variability or fewer SSCs in the joint trajectories. Some recent works use the term “kinematic smoothness” which can be quantified by metrics like number of zero-crossings of acceleration (as seen in a PLOS study: NUM = number of acceleration zero-crossings) (Wattananon et al., 2025) – this is analogous to SSC but for acceleration.

To illustrate application: Suppose we have two conditions – squatting with and without an exoskeleton. Motion capture yields knee angle vs time for each squat. We could compute: Knee ROM in each condition (maybe it stays the same, say $\sim 100^\circ$).

RMS of knee angular velocity in each condition. Perhaps with the exoskeleton, the subject moves slightly slower and more deliberately (due to feeling the device's presence), yielding a lower RMS velocity.

Motion capture also critically allows synchronization with EMG and force data. By combining these, researchers can correlate muscle activation (EMG features from Section 2.4) with joint kinematics. For instance, they might observe that with an exoskeleton, quadriceps EMG RMS is reduced (suggesting assistance), and concurrently the knee angular acceleration is reduced or smoother (suggesting the device took on some of the work and the movement was more controlled). Thus,

kinematic features and EMG features together provide a fuller picture: EMG shows what muscles are doing, MoCap shows what the joints are doing, and any exoskeleton effect should ideally appear in both (reduced muscle effort, maintained or improved kinematics).

In conclusion, 3D motion capture in squat biomechanics enables accurate measurement of joint angles and derived metrics, facilitating detailed analysis of movement patterns (Hoogenboom et al., 2023). Key kinematic features like ROM are used to describe the extent of movement, while additional measures (RMS, MAV of velocities, waveform length of angle signals, etc.) can describe the smoothness and intensity of the movement. These features, although used less frequently than EMG features, are valuable for quantifying differences in technique or the effect of interventions. With the increasing availability of markerless motion capture and wearable IMUs, such kinematic analyses are becoming more accessible, allowing even field assessments of squatting form. In the context of our thesis, motion capture analysis ensures that any assistance provided by an exoskeleton does not detrimentally alter the natural squat kinematics; rather, it should preserve normal ROM and perhaps even reduce unwanted movements (like wobble or asymmetry) by virtue of reducing fatigue. Kinematic features will be examined alongside

EMG features to comprehensively evaluate the exoskeleton's impact on the squatting movement (further detailed in Chapter 3 methodologies and Chapter 4 results).

2.6 Synthesis Of Literature And Current Gaps

This literature review has explored the biomechanics of squatting, the roles of key muscles, actuator technologies for exoskeleton assistance, and the analytical methods (EMG and motion capture) used to study these systems. In synthesizing the findings, several common themes and gaps in current research emerge:

Human-Exoskeleton Integration: Effective squat assistance requires harmonizing with human biomechanics. Prior studies underscore the importance of designing exoskeletons that align with natural joint motion and muscle function (Hoogenboom et al., 2023) (Li et al., 2025). Yet, a gap remains in understanding *long-term adaptation*: how do users adjust their muscle coordination over time when squatting with assistance? Most studies are short-term and indicate immediate muscle activity reductions (Li et al., 2025), but it's unclear if new movement patterns (good or bad) develop with prolonged use. Future research should investigate motor learning effects – for instance, does reliance on an exoskeleton lead to any muscle weakening or changed recruitment strategies when the device is removed? Ensuring that assistance does not foster detrimental motor adaptations is an open question.

Multi-Muscle Coordination and Control: The literature emphasizes that multiple muscles (quads, glutes, hamstrings, etc.) work in synergy during squats (Ottinger et al., 2020) (Physiopedia contributors). However, current exoskeleton control strategies often provide assistance in a simplified manner (e.g., only at one joint or based on a pre-set torque profile). There is a research gap in developing intelligent, multi-modal control that accounts for the user's overall neuromuscular state. Recent trends suggest integrating multimodal sensing – combining EMG signals, motion data, perhaps pressure sensors – to infer the user's intention and fatigue state in real time (Li et al., 2025). For example, an exoskeleton that detects an oncoming muscle fatigue (via frequency shifts in EMG) could adjust its assistance to compensate. Such adaptive, context-aware control is still in early stages.

Actuator Design Trade-offs: While advances in actuator tech (like QDD motors and PAMs) have been substantial, the gap between laboratory prototypes and real-world deployment is notable. One challenge is power and energy: electric exoskeletons with low gear ratios need high torque motors which can strain batteries, whereas pneumatic/hydraulic need portable power sources. Future research is directed toward making actuators lighter and more energy-efficient without sacrificing compliance (Li et al., 2025). For instance, developing lightweight power sources or energy-recycling systems (maybe regenerating energy when descending in a squat) could greatly enhance practical use. Additionally, reliability and durability of these actuators in real-world conditions (dust, temperature changes, long duty cycles) remains a concern – pointing to a need for studies on robustness and maintenance of exoskeleton actuators over time (Li et al., 2025).

User Safety Standards: The importance of safety has been reiterated through all topics. Yet, there is a gap in formalizing safety testing and standards specifically for squat-assist exoskeletons. Researchers note that standard protocols for risk assessment and validation are needed (Li et al., 2025). Future work should involve systematic evaluation of potential failure modes (e.g., what happens if a sensor fails mid-squat, or if the user stumbles) and creating certification processes. Currently, much of the evidence comes from controlled lab studies; translating that into certified safe devices for workplaces or clinics will require bridging engineering with regulatory science (Li et al., 2025).

Biomechanical and Clinical Outcomes: Many studies focus on immediate biomechanical metrics (EMG reductions, reduced joint load (Li et al., 2025), etc.). A gap exists in linking these to actual *clinical or performance outcomes*. Do assisted squats lead to lower injury rates in workers over months or years? Can patients in rehab recover faster or with less pain if they use an assistive device for squats? These long-term and outcome-based evaluations are scarce. Similarly, while some studies have looked at metabolic cost reductions (Jeong et al., 2023), few have addressed whether exoskeleton use might alter things like muscle fiber type engagement or cardiovascular responses in the long run. Such questions would likely require interdisciplinary studies combining biomechanics with physiology.

Motion Capture and EMG Integration: On the methodological side, a gap remains in fully integrating motion and muscle data for real-time control. We have sophisticated analysis tools (as reviewed in Sections 2.4 and 2.5), but using that data in closed-loop exoskeleton control (HMI – human-machine interface) is still being refined (Li et al., 2025). Studies like the one by Chen et al. (2023) indicate growing interest in using EMG for direct control inputs (Li et al., 2025), but reliability and intuitiveness of such interfaces in dynamic tasks like squatting are ongoing challenges. Future research could improve sensor fusion algorithms that reliably interpret a combination of EMG intensity (user effort) and kinematic cues (movement phase detection from MoCap or IMUs) to modulate assistance seamlessly. Achieving this will help exoskeletons feel more *transparent* – responding to the user’s needs without manual mode switches or noticeable delays.

Population Diversity: Most squat biomechanics and exoskeleton studies are done on young healthy males in labs. There is a gap in understanding how these devices work for diverse populations – e.g., older adults, females (who may have different strength ratios or knee valgus tendencies), or individuals with orthopedic conditions. Because muscle activation patterns can differ (e.g., those with knee osteoarthritis might have different activation or kinematics), the assistive strategies might need tailoring. Research into personalized assistance is emerging, where assistance levels or timing are optimized to an individual’s physiology (Jeong et al., 2023). However, more work is needed to determine if one-size-fits-all control is sufficient or if exoskeletons should adapt to specific user profiles (learning algorithms could play a role here).

Chapter 3

Materials and Methods

3.1 Participant Selection And Study Design

This study employed a within-subject experimental design to assess the biomechanical and neuromuscular effects of a hip exoskeleton during bodyweight squatting. The primary objective was to compare muscle activation and joint kinematics during squats performed with and without exoskeletal assistance, with a specific focus on the ascending phase.

A total of 17 healthy male participants were recruited for this study. All subjects were free from lower limb musculoskeletal disorders, neurological conditions, or previous hip or knee surgeries that might interfere with squat performance or data collection. The average participant height was approximately 175 cm, and body weight ranged between 70 and 80 kg, representing a relatively homogeneous sample in terms of body structure and load characteristics.

3.2 Instrumentation And Devices

3.2.1 Electromyography. Surface electromyography (sEMG) signals were captured using the Trigno® Wireless Biofeedback System (Serial No. SP-W02C-1357), widely regarded as a benchmark for sEMG data collection (Jayaraman et al., 2024). Each sensor in the system combines a three-axis accelerometer with an EMG electrode, supporting wireless data transmission over distances up to 40 meters. With a battery life of approximately seven hours and a high sampling rate of 2148 Hz, the Trigno system works seamlessly with Delsys' proprietary EMGworks Acquisition software, facilitating efficient and accurate biological signal acquisition. Its performance is further enhanced by noise-resistant design and specialized parallel bar electrode technology, which together help ensure high signal fidelity (Nair, Rajapriya, & Rajeswari, 2021). Figure (). Trigno wireless sEMG system (Gage, 2023).



Figure 7. Trigno wireless sEMG system (gage, 2023).

3.2.2 Motion capture. To track joint angles and full-body kinematics, the XSENS Awinda system (XSENS Technologies B.V., Enschede, The Netherlands) was employed. This wireless motion capture setup consists of 17 inertial measurement units (IMUs) that communicate via Wi-Fi with a receiver connected to a computer. Operating at a sampling frequency of 60 Hz, the system processes data through the MVN Analyze software to generate a detailed biomechanical model with 23 segments and 22 joints. Each joint is characterized by 3D rotational data across three anatomical planes. The model adheres to the International Society of Biomechanics (ISB) standards for kinematic reporting (Wu et al., 2002).

To ensure synchronized data collection between the XSENS system, the Bertec force plate, and the Trigno® Wireless Biofeedback System (SN: SP-W02C-1357), a Delsys Trigger Module was used. This module receives a trigger pulse from the XSENS base station and relays it to both the AM6800 amplifier and the Trigno EMG base station (SP-W02), all managed through the EMGworks Acquisition software to maintain precise timing across all devices.



Figure 8. Trigno® wireless biofeedback (SN: SP-W02C-1357) System (Gage, 2020).

3.2.3 Powered hip support brace. The exoskeleton developed in this thesis is designed to assist hip extension during squat movements. It comprises a commercial hip brace and a custom-built actuator assembly integrated at the hip joint. The hip brace provides a secure attachment to the user's body and serves as the support structure, while the actuator delivers targeted assistive torque at the hip. Additionally, a sensing and control subsystem is implemented to monitor the hip angle and drive the actuator in real time.



Figure 9. The powered hip support brace

3.3 Data Acquisition

3.3.1 Electromyography. Prior to starting the experimental trials, surface electromyography (sEMG) electrodes were applied to the target lower limb muscles (see Figure) following the placement protocols recommended by SENIAM.org. To ensure optimal signal quality, the selected skin areas were thoroughly cleaned with alcohol and shaved to reduce impedance and eliminate interference (Konrad, 2005). The electrodes were then securely fixed in place using adhesive tape to minimize movement artifacts and maintain consistent contact throughout data collection.

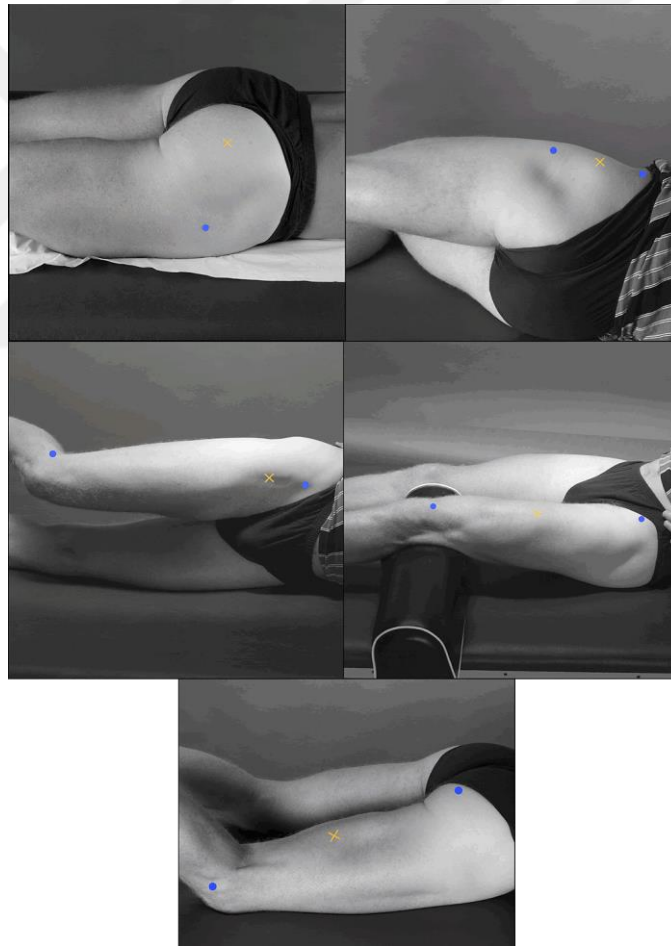


Figure 10. Lower limb EMG sensor placement by SENIAM.

3.3.2 Motion capture. Participants were fitted with the XSENS Awinda motion capture system using a Lycra shirt, Velcro straps, and a headband (Figure 11). Sensor placement strictly adhered to the official guidelines provided by XSENS Movella to ensure accurate readings and minimize noise caused by motion artifacts. After the sensors were in place, a series of body segment measurements were taken—including head and shoulder height, shoulder and elbow width, wrist and arm span, hip and knee height, ankle height, foot length, and sole thickness.

Calibration began with participants standing in a neutral N-pose for 3 seconds, followed by a 16-second walk to and from the starting point. The calibration process was repeated as needed until the MVN software confirmed successful alignment.



Figure 11. Movella xsens motion capture system sensor placement.

3.4 Exoskeleton System Design

3.4.1 Hip brace and actuator design. The hip brace used in this system is a commercial off-the-shelf orthopedic hip brace (WingMed Hip Brace) typically intended for post-operative support. This brace consists of a waistband component and a thigh cuff, connected by an adjustable hinge at the hip joint. It is constructed from rigid thermoplastic material with padded contact surfaces for patient comfort. Importantly, the brace was used without any structural modification, preserving its original strength and alignment features. The hinge mechanism of the brace normally allows the range of hip flexion to be limited (e.g. lockable in 15° increments up to 120°), but for this application it was left unrestricted to allow the full range of motion required for squatting. The brace provides a stable framework aligned with the anatomical hip axis, making it an ideal mounting platform for the actuator without compromising user safety or comfort.

Attached to the brace's lateral hip joint is the custom actuator assembly, which was designed and fabricated specifically for this exoskeleton. The actuator follows a quasi-direct-drive design philosophy (inspired by the open-source OpenTorque actuator) to achieve high torque output with minimal gear reduction. The actuator's housing was 3D-printed in ABS plastic, chosen for its durability and the ability to produce complex, lightweight structures. Inside this housing, a high-torque brushless DC motor drives a compact planetary gear train with a reduction ratio of 9:1 (slightly higher than the 8:1 ratio used in the original OpenTorque design). This moderate gear reduction amplifies the motor's torque while maintaining a relatively low output impedance and partial backdrivability – meaning the joint can still be moved by the user when the motor is unpowered with only minimal resistance. The low gear ratio thus allows the actuator to provide meaningful assistance during lifting movements without significantly hindering the user's natural motion during other phases of the squat.

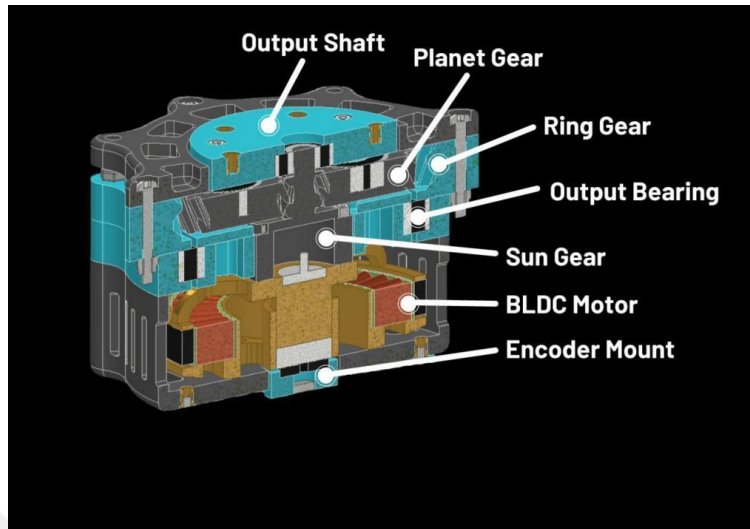


Figure 12. Actuator design inspired by the open-source OpenTorque actuator.

The motor used in the actuator is an Eaglepower 8308 brushless outrunner (205 kV), a powerful motor commonly employed in heavy-lift drones. Its large diameter and high torque constant make it well-suited to produce the forces needed for hip extension support. The motor is coupled to the output through the 9:1 gearbox, which is fully integrated into the actuator assembly. The output shaft of the gearbox is aligned coaxially with the brace's hinge joint, effectively augmenting the brace's passive hip joint with an active, powered rotation axis. One end of the actuator housing is fixed to the brace's waist section, while the gearbox output is linked to the thigh section of the brace. In this way, the actuator can apply assistive torque to rotate the thigh relative to the pelvis about the hip joint. All mounting interfaces were custom-designed to clamp securely onto the existing brace structure without altering it, ensuring that the actuator's forces are transmitted effectively through the brace to the user while the brace continues to provide its normal support and alignment functions.



Figure 13. Eaglepower 8308 brushless outrunner.

3.4.2 Control and sensing. Control of the exoskeleton is managed by an embedded microcontroller (Arduino Uno) that coordinates motor activation based on sensor feedback. Joint angle feedback is provided by a magnetic rotary encoder integrated into the actuator. In this design, a small permanent magnet is attached to a rotating component of the actuator (coaxial with the motor shaft), and a magneto resistive sensor chip is fixed in the housing to measure the magnet's angular position. This non-contact magnetic encoder yields real-time absolute angle measurements of the hip joint, allowing the controller to accurately determine the degree of hip flexion or extension at any moment.

The Arduino microcontroller reads the encoder data and outputs control signals to a brushless DC motor driver (model ZS-X11F) which powers the actuator. The motor driver accepts a pulse-width modulated (PWM) signal for speed control and a direction signal from the Arduino to regulate the motor's rotation. Using the real-time angle feedback, a simple state-based control algorithm was implemented to provide assistance only when appropriate during the squat. Specifically, as the user descends into a squat (increasing hip flexion angle), the system remains passive – the motor is kept off, allowing the brace and joint to move freely. Once the hip joint passes a predetermined flexion angle threshold (approximately 70° of flexion), the controller “arms” the assistive mode. Then, as the user begins to ascend from the squat (hip angle decreasing from that peak flexion), the Arduino activates the motor driver to generate torque in the hip extension direction. The actuator thus actively assists by driving the hip joint from the flexed position back

toward full extension (0° flexion). To ensure safety and smoothness, the motor's output is ramped up gradually via the PWM control rather than applying full power instantly, preventing any jerky motion. When the hip approaches full extension again (or if an external limit switch or safety stop is triggered), the controller cuts power to the motor and engages a braking or idle mode. This completes one assistance cycle, after which the system resets its logic and awaits the next repetition.

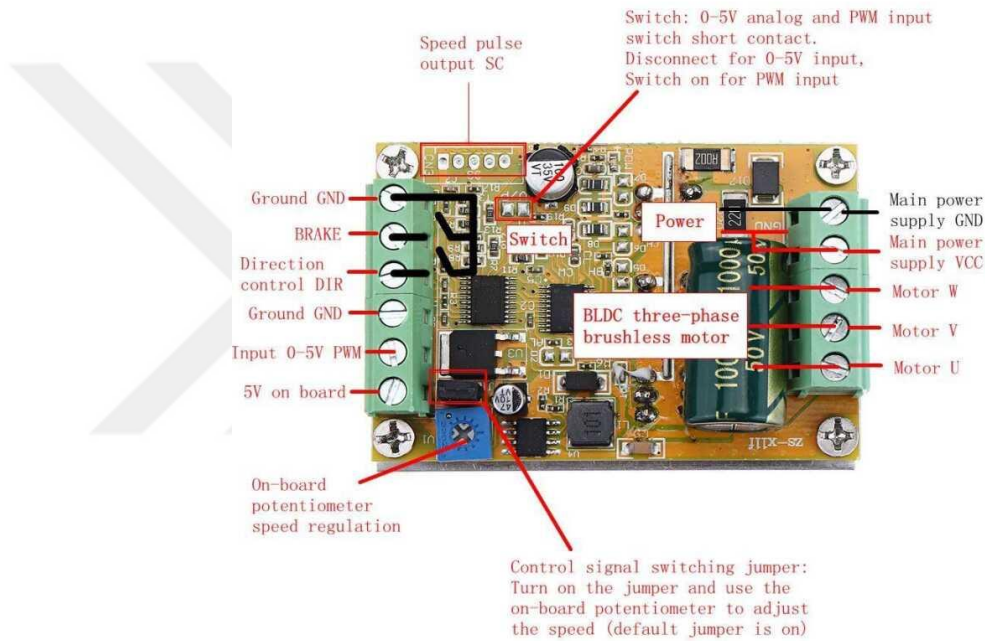


Figure 14. Brushless DC motor driver (model ZS-X11F)

Overall, this exoskeleton system design combines the sturdy structural support of a medical-grade hip brace with the active power of a modular actuator and a responsive control scheme. The integration is accomplished in a way that maintains user comfort and safety: the brace handles load-bearing and alignment, while the actuator provides on-demand assistance only when needed. The modular nature of the design (using a clip-on actuator with an unmodified brace) means that the device can be attached or removed with relative ease, and the underlying brace can still be used in a conventional passive manner when the actuator is not in operation. This approach to design ensured that the hip exoskeleton could augment the user's strength during strenuous movements like rising

from a squat, without impeding natural joint function or range of motion during other activities.

3.5 Experimental Protocol

To evaluate the biomechanical and neuromuscular impact of the hip exoskeleton system, participants were instructed to perform a series of controlled squatting exercises under two distinct conditions: without the exoskeleton (baseline) and with the exoskeleton activated (assisted). Each participant completed five repetitions of bodyweight squats per condition.

The squatting tasks were performed on a flat, non-slip surface with the participants maintaining a consistent foot stance approximately shoulder-width apart. A standardized verbal pacing cue was provided to ensure uniform tempo across trials and between participants. Each squat followed a controlled cycle: a smooth descent phase into the squat, followed by an ascent back to an upright position. The ascending phase, during which the actuator was configured to provide assistance, was the primary focus of data analysis.

To minimize the effects of fatigue and ensure repeatability, participants were given a mandatory rest interval of five minutes between the two squat conditions. This recovery period was intended to prevent muscular fatigue from influencing muscle activation or joint mechanics during the assisted trials.

No external weights or loads were applied during any phase of the experiment; only the subject's body weight was used to preserve the natural biomechanics of the movement and isolate the influence of the hip exoskeleton. Each participant completed the entire protocol within a single testing session. The order of the conditions (brace first or no brace first) was randomized to avoid any systematic bias due to learning or fatigue effects.

Throughout the squatting tasks, electromyographic and kinematic data were simultaneously recorded to capture muscular activation and joint behavior in both assisted and unassisted conditions. These data were used for the comparative analysis of muscle

load reduction and joint kinematic adaptation induced by the hip exoskeleton during the ascending phase of the squat.

3.6 Data Analysis

To obtain accurate data suitable for in-depth analysis, preprocessing techniques such as detrending, rectification, filtering, and smoothing were applied to the raw sEMG signals to remove noise and bias. A fourth-order Butterworth bandpass filter with cutoff frequencies of 20 Hz and 450 Hz was used for filtering, along with a notch filter at 50 Hz to reduce power line interference.

The data obtained from the sEMG and IMU-based motion capture systems initially possessed different sampling rates. Consequently, an up-sampling using bicubic interpolation step was applied to the joint kinematic data to match the sampling frequency of the sEMG datasets.

Squat phase segmentation was performed to analyze the full squat cycle, as well as its distinct descent and ascent phases. Individual squat cycles were manually segmented using the `ginput` function in MATLAB, with the knee flexion/extension angle serving as a visual reference to ensure accurate demarcation, addressing inter-subject performance variability. Within each segmented squat cycle, the descent phase (initiation of knee flexion to peak knee flexion) and ascent phase (peak knee flexion to full knee extension) were identified using a peak detection algorithm applied to the knee flexion/extension angle.

Subsequently, both sEMG and joint kinematic data for each identified phase (full squat, descent, ascent) were time-normalized to 100 data points, representing 0-100% of the phase duration. A detailed analysis and calculation procedure was carried out on the processed data to extract relevant parameters and features using custom scripts in MATLAB (The MathWorks, Inc., Natick, MA, USA). The following features were extracted:

sEMG Features:

- Root Mean Square (RMS)

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

where x_i is the i -th sample of the EMG signal and N is the total number of samples.

- Mean Absolute Value (MAV)

$$\text{MAV} = \left(\frac{1}{N} \right) \sum_{i=1}^N |x_i| \quad (2)$$

where x_i is the i -th sample of the EMG signal and N is the total number of samples.

- Area Under the Curve (AUC)

$$\text{AUC} = \left(\sum_{i=1}^N x_i \right) \cdot \Delta t \quad (3)$$

where x_i is the i -th sample of the sEMG signal, N is the total number of samples, and Δt is the time interval between samples.

- Variance (VAR)

$$\text{VAR} = \frac{1}{N-1} \sum_{i=1}^N (x_i - \bar{x})^2 \quad (4)$$

where x_i is the i -th sample of the sEMG signal, N is the total number of samples, and $\bar{x}$ is the mean of the sEMG signal over the N samples

Kinematic Features:

- Range of Motion (ROM)

$$\text{ROM} = \max(\theta) - \min(\theta) \quad (5)$$

where θ is the joint angle over the squat cycle.

- Root Mean Square (RMS)

$$\text{RMS} = \sqrt{\left[\left(\frac{1}{N} \right) \sum_{i=1}^N \theta_i^2 \right]} \quad (6)$$

where θ_i is the joint angle at the i -th sample and N is the total number of samples.

- Mean Absolute Value

$$\text{MAVAngle} = \left(\frac{1}{N} \right) \sum_{i=1}^N |\theta_i| \quad (7)$$

where θ_i is the joint angle at the i -th sample and N is the total number of samples.

These features were calculated for both "without brace" and "with brace" conditions for the full squat, descent, and ascent phases.

3.7 Statistical Analysis

All statistical analyses were performed using MATLAB. Prior to performing statistical tests, the normality of the paired differences (With Brace - Without Brace) for each feature was assessed using the Shapiro-Wilk test. For data that followed a normal distribution, a parametric paired t-test was employed to determine statistically significant changes in the EMG and kinematic features between the "without brace" and "with brace" conditions. For non-normally distributed data, the Wilcoxon signed-rank test, a non-parametric alternative, was applied.

Statistical significance was set at $p < 0.05$, and all analyses were conducted as two-tailed tests. For features that showed a statistically significant difference, the percentage change was calculated using the "without brace" condition as the baseline:

$$\text{Percentage of Change (\%)} = \left(\frac{\text{Feature}_{\text{with brace}} - \text{Feature}_{\text{without brace}}}{\text{Feature}_{\text{without brace}}} \right) \times 100 \quad (8)$$

Only statistically significant percentage changes were reported in the results.

Furthermore, a correlation analysis was conducted on the kinematic data to investigate the relationship between changes in hip flexion/extension and knee flexion/extension kinematics due to the brace. This analysis utilized Spearman's rank correlation coefficient (ρ). The correlation was performed on the difference values: $\Delta\text{Feature} = \text{Feature}_{\text{with brace}} - \text{Feature}_{\text{without brace}}$ (9) for each participant. This correlation was only applied to pairs of kinematic features ($\Delta\text{ROMHipFlexExt}$ VS $\Delta\text{ROMHipFlexExt}$) where both respective features (hip flexion/extension feature and knee flexion/extension feature) had demonstrated a statistically significant change between the brace conditions from the paired tests.

This entire data analysis and statistical procedure was independently applied to the data derived from the full squat cycle, the descent phase, and the ascent phase.

Chapter 4

Results

4.1 Full Squat Cycle Analysis

4.1.1 Visual overview of the time-normalized data profiles. To provide a qualitative overview of the muscle activation patterns and joint kinematics throughout the full squat cycle, ensemble averaged, time normalized, profiles were generated for both the 'with brace' and 'without brace' conditions. Figure 15 presents the mean ($\pm$ standard deviation, SD) time-normalized sEMG profiles for the monitored muscles. Figure 16 displays the mean ($\pm$ SD) time-normalized joint angle profiles for the analyzed lower limb joint movements. These profiles illustrate the general temporal characteristics of muscle activity and joint motion across the normalized squat cycle (0-100 frames).

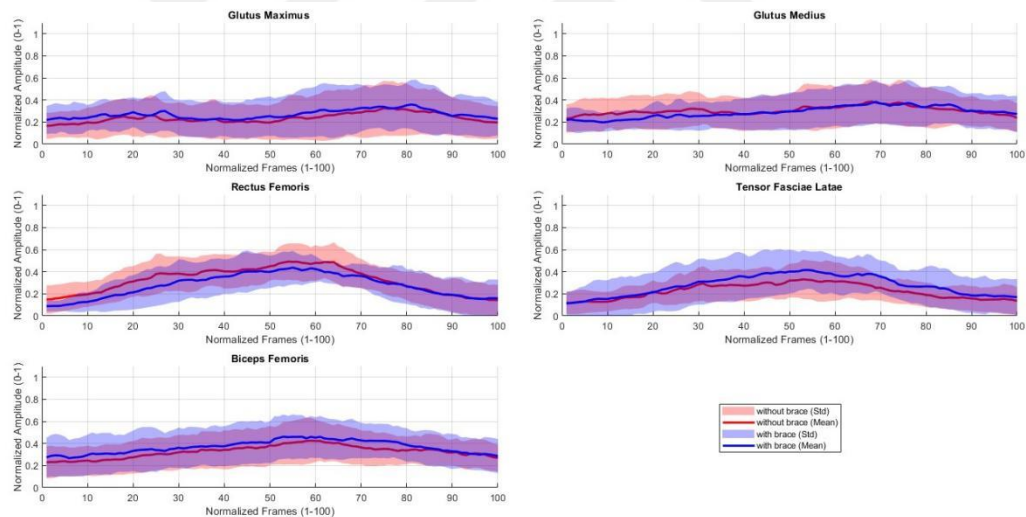


Figure 15. Mean ($\pm$ standard deviation) time/amplitude normalized sEMG amplitude profiles for Gluteus Maximus, Gluteus Medius, Rectus Femoris, Tensor Fasciae Latae, and Biceps Femoris during the full squat cycle. Data are shown for conditions with the hip brace (blue lines and shading) and without the hip brace (red lines and shading).

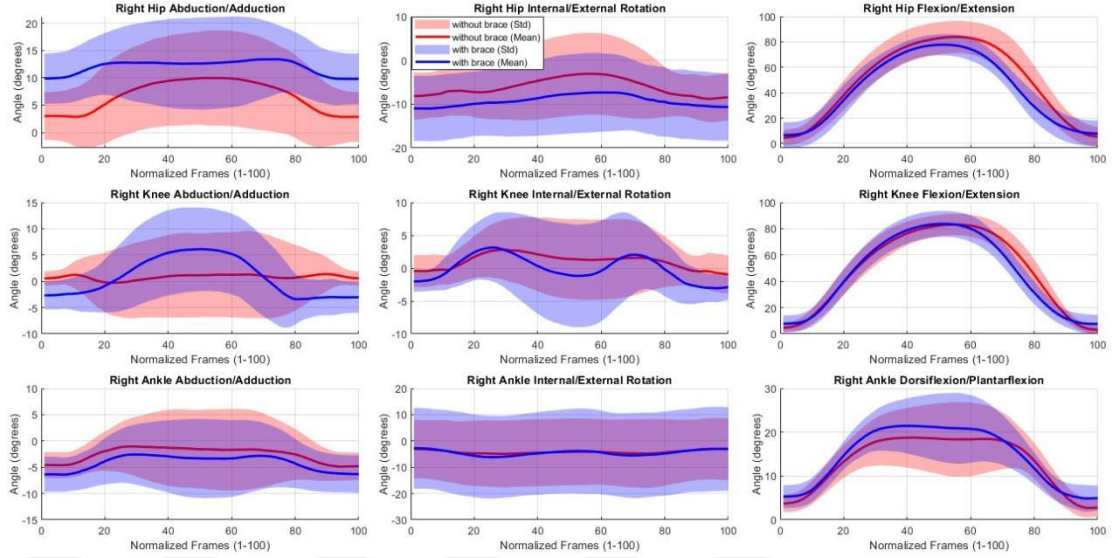


Figure 16. Mean ($\pm$ standard deviation) time-normalized joint angle profiles. data are shown for conditions with the hip brace (blue lines and shading) and without the hip brace (red lines and shading).

4.1.2 Electromyography results – full squat. Table1 summarizes the descriptive statistics [Mean (SD)] and p-values for the linear sEMG features (RMS, MAV, AUC, VAR) recorded from the Biceps Femoris, Gluteus Maximus, Gluteus Medius, Rectus Femoris, and Tensor Fasciae Latae during the full squat cycle, comparing the "with brace" condition to the "without brace" condition.

Table 1

Descriptive Statistics Mean (SD) And Statistical Comparison (P-Values) Of sEMG Features (RMS, MAV, AUC, VAR) For Selected Lower Limb Muscles During The Full Squat Cycle Performed With And Without The Hip Brace

	without brace Mean (SD)	with brace Mean (SD)	P-value
Biceps Femoris			
RMS	3.49E-01 (1.22E-01)	3.81E-01 (1.24E-01)	0.231
MAV	3.19E-01 (1.23E-01)	3.43E-01 (1.39E-01)	0.687
AUC	1.49E-02 (5.72E-03)	1.60E-02 (6.45E-03)	0.687
VAR	2.03E-02 (1.03E-02)	2.42E-02 (1.16E-02)	0.137
Gluteus Maximus			
RMS	2.65E-01 (1.49E-01)	2.89E-01 (1.42E-01)	0.634
MAV	2.29E-01 (1.60E-01)	2.54E-01 (1.55E-01)	0.630
AUC	1.07E-02 (7.43E-03)	1.18E-02 (7.24E-03)	0.630

Table 1 (cont.d)

VAR	1.61E-02 (6.96E-03)	1.57E-02 (4.95E-03)	0.795
Gluteus Medius			
RMS	3.19E-01 (1.25E-01)	2.92E-01 (1.40E-01)	0.229
MAV	2.91E-01 (1.31E-01)	2.64E-01 (1.42E-01)	0.236
AUC	1.35E-02 (6.09E-03)	1.23E-02 (6.59E-03)	0.236
VAR	1.69E-02 (6.83E-03)	1.62E-02 (9.08E-03)	0.704
Rectus Femoris			
RMS	3.53E-01 (7.42E-02)	2.98E-01 (6.95E-02)	0.019
MAV	2.93E-01 (8.28E-02)	2.32E-01 (6.20E-02)	0.008
AUC	1.36E-02 (3.85E-03)	1.08E-02 (2.88E-03)	0.008
VAR	3.80E-02 (1.28E-02)	3.69E-02 (1.55E-02)	0.815
Tensor Fasciae Latae			
RMS	2.42E-01 (1.01E-01)	2.99E-01 (1.04E-01)	0.186
MAV	1.95E-01 (9.18E-02)	2.37E-01 (1.08E-01)	0.314
AUC	9.07E-03 (4.27E-03)	1.10E-02 (5.04E-03)	0.314
VAR	2.33E-02 (1.41E-02)	3.35E-02 (1.48E-02)	0.033

The analysis of sEMG features comparing the "with brace" to "without brace" conditions during the full squat cycle revealed statistically significant percentage changes for specific muscles and features, as detailed in Table 2

Table 2

Percentage Change (%) In sEMG Features For Selected Lower Limb Muscles When Wearing The Hip Brace Compared To The No-Brace Condition During The Full Squat Cycle

Features	Biceps Femoris	Gluteus Maximus	Gluteus Medius	Rectus Femoris	Tensor Fasciae Latae
RMS	NS	NS	NS	-15.37%	NS
MAV	NS	NS	NS	-20.84%	NS
AUC	NS	NS	NS	-20.84%	NS
VAR	NS	NS	NS	NS	43.70%

NS indicates no significant change.

- Biceps Femoris: No statistically significant changes were observed in any of the analyzed sEMG features (RMS, MAV, AUC, VAR) for the Biceps Femoris when wearing the hip brace compared to the no-brace condition (all $p > 0.05$). (Table 1)

- Gluteus Maximus: Similar to the Biceps Femoris, the Gluteus Maximus showed no statistically significant alterations in RMS, MAV, AUC, or VAR activity during the full squat cycle with the use of the hip brace (all $p > 0.05$). (Table 1)
- Gluteus Medius: The activity of the Gluteus Medius, as measured by RMS, MAV, AUC, and VAR, was not significantly affected by the hip brace during the full squat cycle (all $p > 0.05$).
- Rectus Femoris: The Rectus Femoris exhibited significant reductions in muscle activity when the hip brace was worn, as illustrated in Figure 17 and table 2. Specifically, RMS decreased by 15.37% ($p = 0.019$), MAV decreased by 20.84% ($p = 0.008$), and AUC also decreased by 20.84% ($p = 0.008$). The VAR feature for the Rectus Femoris did not show a significant change ($p = 0.815$).
- Tensor Fasciae Latae: For the Tensor Fasciae Latae, a statistically significant increase was observed in the VAR feature, which rose by 43.70% ($p = 0.033$) with the hip brace. The RMS, MAV, and AUC features did not show any significant changes for this muscle (all $p > 0.05$).

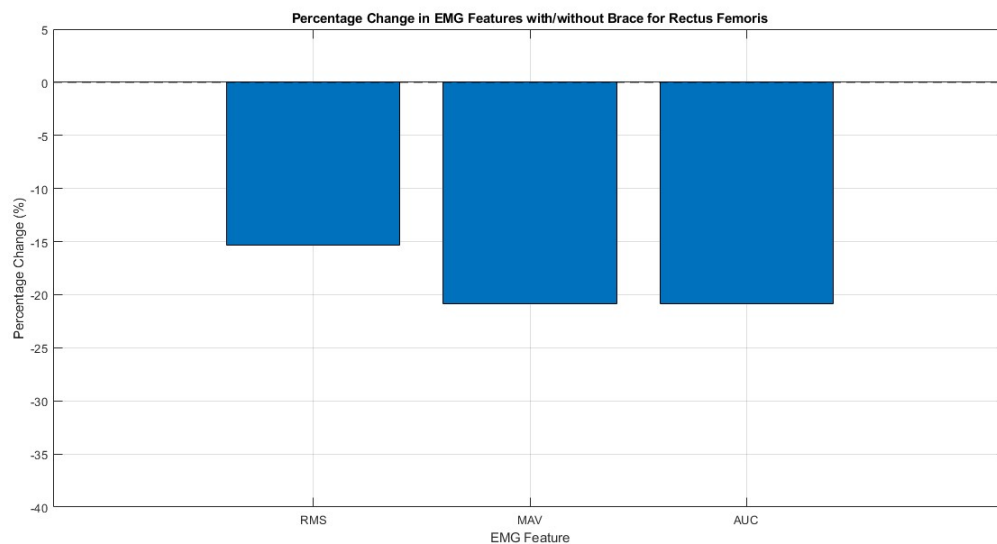


Figure 17. Percentage change in sEMG features (RMS, MAV, AUC) for the rectus femoris muscle during the full squat cycle when using the hip brace. the bars represent the percentage change in EMG features when comparing the ‘with brace’ condition to the ‘without brace’ (baseline) condition. Negative values indicate a reduction in muscle activity with the brace for these features.

4.1.3 Kinematic results – full squat. The descriptive statistics Mean (SD) and p-values for the kinematic features (Range of Motion - ROM, Root Mean Square - RMS, Mean Absolute Value - MAV) for the right hip, knee, and ankle joints during the full squat cycle are presented in Table 3, comparing the "with brace" to the "without brace" conditions.

Table 3

Descriptive Statistics Mean (SD) And Statistical Comparison (P-Values) Of Kinematic Features (ROM, RMS, MAV) For Right Hip, Knee, And Ankle Joint Movements During The Full Squat Cycle Performed With And Without The Hip Brace

Feature	Joint	Without Brace mean (SD)	With Brace mean (SD)	P-value
Flexion/Extension				
ROM	Hip	81.60 (12.80)	73.75 (11.02)	0.008
RMS	Hip	53.42 (9.89)	45.60 (6.10)	0.003
MAV	Hip	43.63 (8.99)	35.89 (7.38)	0.005
ROM	Knee	81.57 (7.69)	79.35 (11.57)	0.320
RMS	Knee	52.98 (6.96)	49.16 (6.88)	0.026
MAV	Knee	42.49 (6.84)	38.29 (6.77)	0.015
ROM	Ankle	19.84 (5.07)	19.95 (5.14)	0.908
RMS	Ankle	13.89 (3.86)	14.61 (3.36)	0.279
MAV	Ankle	11.90 (3.23)	12.48 (2.99)	0.351
Abduction/Adduction				
ROM	Hip	10.37 (4.16)	9.90 (3.88)	0.614
RMS	Hip	8.21 (5.22)	12.26 (5.13)	0.000
MAV	Hip	7.45 (5.02)	11.66 (5.16)	0.000
ROM	Knee	9.36 (4.14)	14.13 (5.50)	0.001
RMS	Knee	4.78 (3.23)	5.90 (3.01)	0.068
MAV	Knee	3.96 (2.71)	4.85 (2.43)	0.099
ROM	Ankle	8.19 (4.20)	8.53 (3.47)	0.587
RMS	Ankle	5.44 (2.71)	6.59 (2.87)	0.042
MAV	Ankle	4.95 (2.54)	6.07 (2.85)	0.084
Internal/External Rotation				
ROM	Hip	10.70 (7.17)	8.37 (5.67)	0.031
RMS	Hip	8.69 (4.23)	10.43 (6.96)	0.342
MAV	Hip	8.12 (4.32)	10.01 (7.05)	0.303
ROM	Knee	9.32 (4.13)	12.34 (4.04)	0.012
RMS	Knee	4.00 (2.12)	4.74 (2.52)	0.117
MAV	Knee	3.39 (1.89)	4.03 (2.17)	0.102
ROM	Ankle	7.73 (3.44)	8.23 (2.58)	0.540
RMS	Ankle	8.06 (9.91)	10.69 (12.11)	0.087

Table 3 (cont.d)

MAV	Ankle	7.68 (9.98)	10.29 (12.24)	0.093
-----	-------	-------------	---------------	-------

Table 4 details the statistically significant percentage changes in these kinematic features when the hip brace was worn. A visual summary of these percentage changes across all kinematic features is provided in the heatmap (Figure 18).

Table 4

Percentage Change (%) In Kinematic Features For Right Hip, Knee, And Ankle Joint Movements When Wearing The Hip Brace Compared To The No-Brace Condition During The Full Squat Cycle

Feature	Hip	Knee	Ankle
Flexion/Extension			
ROM	-9.62%	NS	NS
RMS	-14.64%	-7.20%	NS
MAV	-17.74%	-9.88%	NS
Internal/External Rotation			
ROM	-21.81%	32.47%	NS
RMS	NS	NS	NS
MAV	NS	NS	NS
Abduction/Adduction			
ROM	NS	50.96%	NS
RMS	49.31%	NS	21.30%
MAV	56.64%	NS	NS

NS indicates no significant change.

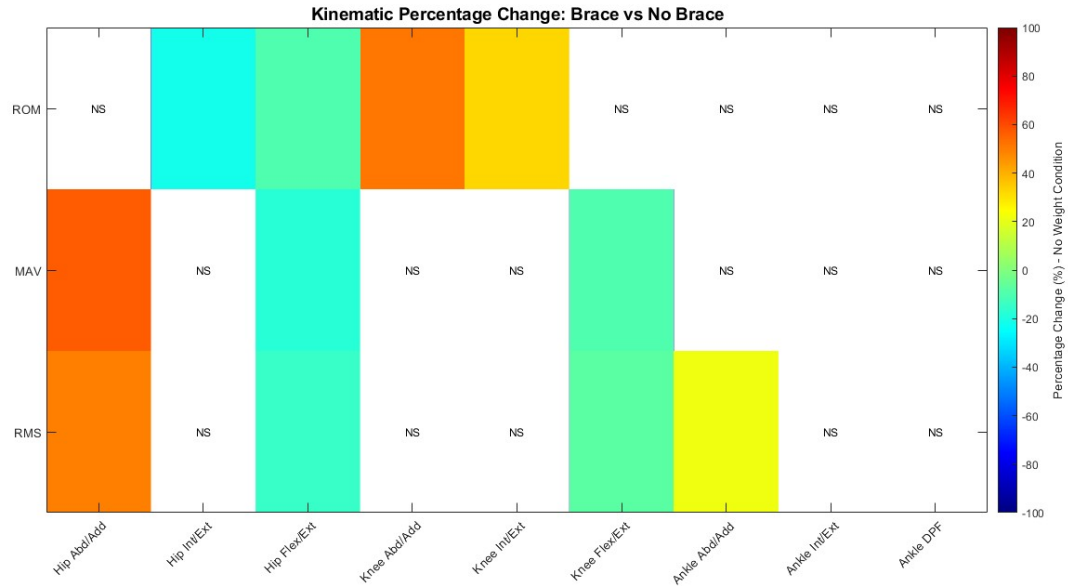


Figure 18. Heatmap illustrating the percentage change in kinematic features (ROM, MAV, RMS) across hip, knee, and ankle joint movements during the full squat cycle when comparing the 'with brace' to the 'without brace' (baseline) conditions. color intensity indicate the magnitude and direction of change (warmer colors for increases, cooler colors for decreases). ns indicates non-significant changes.

Right Hip Joint:

- Flexion/Extension: Wearing the hip brace resulted in significant decreases in hip flexion/extension. ROM decreased by 9.62% ($p = 0.008$), RMS (angle variability) decreased by 14.64% ($p = 0.003$), and MAV (average angle) decreased by 17.74% ($p = 0.005$).
- Abduction/Adduction: For hip abduction/adduction, RMS significantly increased by 49.31% ($p = 0.000$), and MAV significantly increased by 56.64% ($p = 0.000$) with the brace. ROM showed no significant change ($p = 0.614$).
- Internal/External Rotation: Hip internal/external rotation ROM significantly decreased by 21.81% ($p = 0.031$) when using the brace. RMS and MAV for this movement were not significantly affected ($p > 0.05$).

Right Knee Joint:

- Flexion/Extension: For knee flexion/extension, the brace led to a significant decrease in RMS by 7.20% ($p = 0.026$) and MAV by 9.88% ($p = 0.015$). The ROM for knee flexion/extension was not significantly altered ($p = 0.320$).
- Abduction/Adduction: Knee abduction/adduction ROM showed a substantial significant increase of 50.96% ($p = 0.001$) with the brace. RMS and MAV for this movement did not change significantly ($p > 0.05$).
- Internal/External Rotation: Knee internal/external rotation ROM significantly increased by 32.47% ($p = 0.012$) with the brace. No significant changes were observed for RMS or MAV ($p > 0.05$).

Right Ankle Joint:

- Dorsiflexion/Plantarflexion (Flexion/Extension): No statistically significant changes were observed in ROM, RMS, or MAV for ankle dorsiflexion/plantarflexion when wearing the hip brace (all $p > 0.05$).
- Abduction/Adduction: Ankle abduction/adduction RMS showed a significant increase of 21.30% ($p = 0.042$) with the brace. ROM and MAV for this movement did not exhibit significant changes ($p > 0.05$).
- Internal/External Rotation: The hip brace did not lead to any statistically significant changes in ROM, RMS, or MAV for ankle internal/external rotation (all $p > 0.05$).

4.1.4 Kinematic correlation analysis – full squat. To investigate the relationships between brace-induced changes in hip and knee kinematics during the full squat cycle, Spearman's rank correlation analysis (ρ) was performed on the change scores (Δ , representing 'with brace' minus 'without brace' values) for hip flexion/extension and knee flexion/extension features that were both significantly altered by the brace in the preceding analysis (as detailed in Section 4.1.3). The results of this correlation analysis are summarized in Table 5, and illustrative scatter plots for significant correlations are presented in Figure 19.

Significant positive correlations were observed: the change (Δ) in hip flexion/extension ROM was positively correlated with the change (Δ) in knee flexion/extension RMS ($\rho = 0.625$, $p = 0.009$) and also with the change (Δ) in knee flexion/extension MAV ($\rho = 0.544$, $p = 0.026$). Furthermore, the change (Δ) in hip flexion/extension RMS showed significant positive correlations with both the change (Δ) in knee flexion/extension RMS ($\rho = 0.515$, $p = 0.037$) and the change (Δ) in knee flexion/extension MAV ($\rho = 0.593$, $p = 0.014$). No significant correlations were found between the change (Δ) in hip flexion/extension MAV and the changes (Δ) in either knee flexion/extension RMS ($p = 0.159$) or MAV ($p = 0.073$).

Table 5

Spearman's Rank Correlation Coefficients (ρ) And P-Values For The Relationship Between Brace-Induced Change Scores (Δ) In Hip Flexion/Extension Features And Knee Flexion/Extension Features During The Full Squat Cycle

Hip Flexion/extension Features	Knee Flexion/extension Features	Rho	P-Value
ROM	RMS	0.625	0.009
ROM	MAV	0.544	0.026
RMS	RMS	0.515	0.037
RMS	MAV	0.593	0.014
MAV	RMS	0.358	0.159
MAV	MAV	0.449	0.073

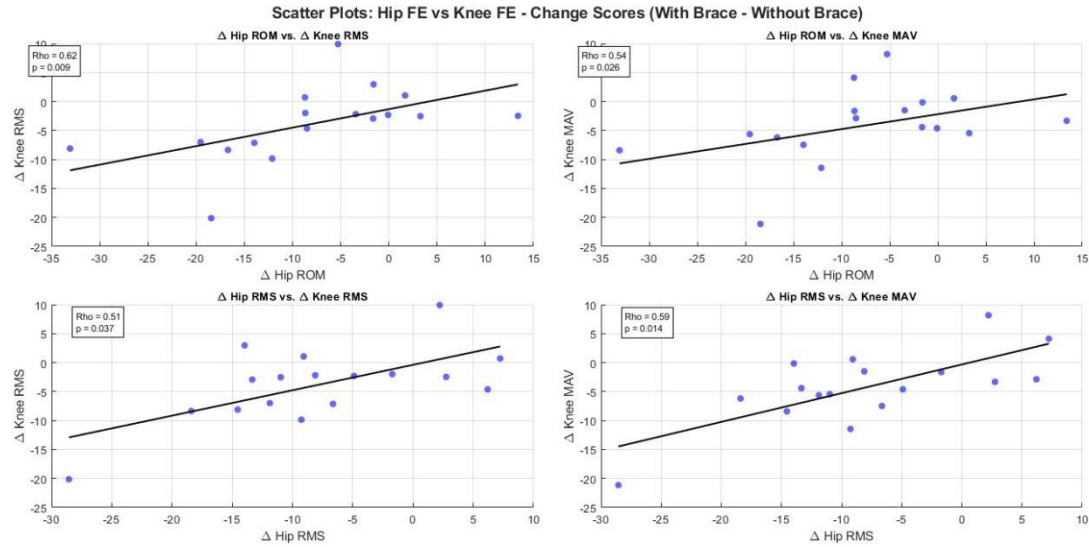


Figure 19. Scatter plots illustrating significant positive correlations between brace-induced change scores (Δ) in selected hip flexion/extension (FE) features and knee flexion/extension (FE) features during the full squat cycle. (a) Δ Hip ROM vs. Δ Knee RMS, (b) Δ Hip ROM vs. Δ Knee MAV, (c) Δ Hip RMS vs. Δ Knee RMS, (d) Δ Hip RMS vs. Δ Knee MAV. Spearman's Rho (ρ) and p-values are shown for each plot.

4.2 Descent Phase Analysis

4.2.1 Electromyography results – descent phase. Table 6 summarizes the descriptive statistics [Mean (SD)] and p-values for the linear sEMG features (RMS, MAV, AUC, VAR) recorded from the Biceps Femoris, Gluteus Maximus, Gluteus Medius, Rectus Femoris, and Tensor Fasciae Latae during the descent phase of the squat, comparing the "with brace" condition to the "without brace" condition.

Table 6

Descriptive Statistics Mean (SD) And Statistical Comparison (P-Values) Of Semg Features (RMS, MAV, AUC, VAR) For Selected Lower Limb Muscles During The Descent Phase Of The Squat Performed With And Without The Hip Brace

	without brace Mean (SD)	with brace Mean (SD)	P-value
Biceps Femoris			
RMS	3.74E-01 (1.21E-01)	4.05E-01 (1.11E-01)	0.287
MAV	3.48E-01 (1.22E-01)	3.66E-01 (1.28E-01)	0.545
AUC	1.62E-02 (5.69E-03)	1.71E-02 (5.98E-03)	0.545
VAR	1.91E-02 (1.24E-02)	2.62E-02 (1.71E-02)	0.097
Gluteus Maximus			
RMS	2.83E-01 (1.86E-01)	3.02E-01 (1.77E-01)	0.746
MAV	2.57E-01 (1.87E-01)	2.76E-01 (1.80E-01)	0.754
AUC	1.20E-02 (8.71E-03)	1.28E-02 (8.38E-03)	0.754
VAR	1.46E-02 (9.92E-03)	1.57E-02 (9.95E-03)	0.751
Gluteus Medius			
RMS	3.37E-01 (1.36E-01)	3.29E-01 (1.64E-01)	0.784
MAV	3.12E-01 (1.39E-01)	3.02E-01 (1.66E-01)	0.752
AUC	1.45E-02 (6.48E-03)	1.41E-02 (7.72E-03)	0.752
VAR	1.66E-02 (1.08E-02)	1.70E-02 (1.00E-02)	0.896
Rectus Femoris			
RMS	3.44E-01 (8.01E-02)	3.17E-01 (7.87E-02)	0.344
MAV	2.86E-01 (9.14E-02)	2.57E-01 (8.28E-02)	0.318
AUC	1.33E-02 (4.25E-03)	1.20E-02 (3.85E-03)	0.318
VAR	3.56E-02 (1.83E-02)	3.45E-02 (1.01E-02)	0.856
Tensor Fasciae Latae			
RMS	2.34E-01 (1.17E-01)	3.14E-01 (1.06E-01)	0.091
MAV	1.97E-01 (1.05E-01)	2.57E-01 (1.13E-01)	0.191
AUC	9.16E-03 (4.89E-03)	1.20E-02 (5.26E-03)	0.191
VAR	1.99E-02 (1.65E-02)	3.24E-02 (1.74E-02)	0.022

Overall, the analysis of sEMG features during the descent phase revealed limited effects of the hip brace on muscle activity. The detailed findings for each muscle are as follows:

- Biceps Femoris: No statistically significant changes were observed in any of the analyzed sEMG features (RMS, MAV, AUC, VAR) for the Biceps Femoris during the descent phase when wearing the hip brace (all $p > 0.05$).

- Gluteus Maximus: The Gluteus Maximus showed no statistically significant alterations in RMS, MAV, AUC, or VAR activity during the descent phase with the use of the hip brace (all $p > 0.05$).
- Gluteus Medius: The activity of the Gluteus Medius, as measured by RMS, MAV, AUC, and VAR, was not significantly affected by the hip brace during the descent phase (all $p > 0.05$).
- Rectus Femoris: Similarly, no statistically significant changes were found in RMS, MAV, AUC, or VAR for the Rectus Femoris during the descent phase when the hip brace was worn (all $p > 0.05$).
- Tensor Fasciae Latae: For the Tensor Fasciae Latae, the RMS, MAV, and AUC features did not show any significant changes during the descent phase (all $p > 0.05$). However, a statistically significant increase was observed in the VAR feature ($p = 0.022$), which rose by 62.81% with the hip brace.

4.2.2 Kinematic results – descent phase. The descriptive statistics Mean (SD) and p-values for the kinematic features (ROM, RMS, MAV) for the right hip, knee, and ankle joints during the descent phase of the squat are presented in Table 7, comparing the "with brace" to the "without brace" conditions.

Table 7

Descriptive Statistics Mean (SD) And Statistical Comparison (P-Values) Of Kinematic Features (ROM, RMS, MAV) For Right Hip, Knee, And Ankle Joint Movements During The Descent Phase Of The Squat Performed With And Without The Hip Brace

Feature	Joint	Without Brace mean (SD)	With Brace mean (SD)	P-value
Flexion/Extension ROM	Hip	80.35 (13.12)	71.90 (10.79)	0.0064

Table 7 (cont.d)

RMS	Hip	52.68 (10.22)	44.71 (6.25)	0.0036
MAV	Hip	42.89 (9.33)	34.95 (7.66)	0.0042
ROM	Knee	81.20 (7.81)	78.07 (11.74)	0.1749
RMS	Knee	51.79 (7.62)	47.61 (6.85)	0.0315
MAV	Knee	40.81 (7.70)	36.58 (6.98)	0.0309
ROM	Ankle	18.61 (5.43)	18.52 (5.66)	0.9353
RMS	Ankle	13.32 (4.13)	13.99 (3.59)	0.3420
MAV	Ankle	11.23 (3.47)	11.90 (3.14)	0.3227
Abduction/Adduction				
ROM	Hip	9.39 (4.44)	9.22 (3.78)	0.8646
RMS	Hip	8.16 (5.35)	12.18 (5.05)	0.0003
MAV	Hip	7.44 (5.14)	11.59 (5.12)	0.0003
ROM	Knee	8.71 (4.25)	13.69 (5.48)	0.0005
RMS	Knee	4.75 (3.25)	5.94 (3.06)	0.1004
MAV	Knee	3.90 (2.68)	4.91 (2.50)	0.0703
ROM	Ankle	6.98 (3.89)	7.56 (3.22)	0.4023
RMS	Ankle	5.53 (2.86)	6.60 (3.00)	0.0552
MAV	Ankle	5.06 (2.75)	6.10 (2.98)	0.0524
Internal/External Rotation				
ROM	Hip	9.35 (7.51)	7.74 (5.27)	0.1899
RMS	Hip	8.72 (4.28)	10.23 (7.08)	0.4110
MAV	Hip	8.19 (4.37)	9.84 (7.15)	0.3726
ROM	Knee	8.49 (4.34)	11.32 (3.91)	0.0186
RMS	Knee	3.84 (1.99)	4.76 (2.56)	0.0670
MAV	Knee	3.27 (1.76)	4.11 (2.22)	0.0444
ROM	Ankle	6.30 (2.81)	6.80 (2.34)	0.5266
RMS	Ankle	7.92 (9.99)	10.69 (12.34)	0.0717
MAV	Ankle	7.56 (10.08)	10.33 (12.47)	0.0772

Table 8 details the statistically significant percentage changes in these kinematic features when the hip brace was worn during the descent phase. A visual summary of these percentage changes across all kinematic features for the descent phase is provided in the heatmap (Figure 20).

Table 8

Percentage Change (%) In Kinematic Features For Right Hip, Knee, And Ankle Joint Movements When Wearing The Hip Brace Compared To The No-Brace Condition

During The Descent Phase Of The Squat. Only Statistically Significant ($P < 0.05$) Percentage Changes.

Feature	Hip	Knee	Ankle
Flexion/Extension			
ROM	-11%	NS	NS
RMS	-15%	-8%	NS
MAV	-19%	-10%	NS
Internal/External Rotation			
ROM	NS	33%	NS
RMS	NS	NS	NS
MAV	NS	26%	NS
Abduction/Adduction			
ROM	NS	57%	NS
RMS	49%	NS	NS
MAV	56%	NS	NS

NS indicates no significant change

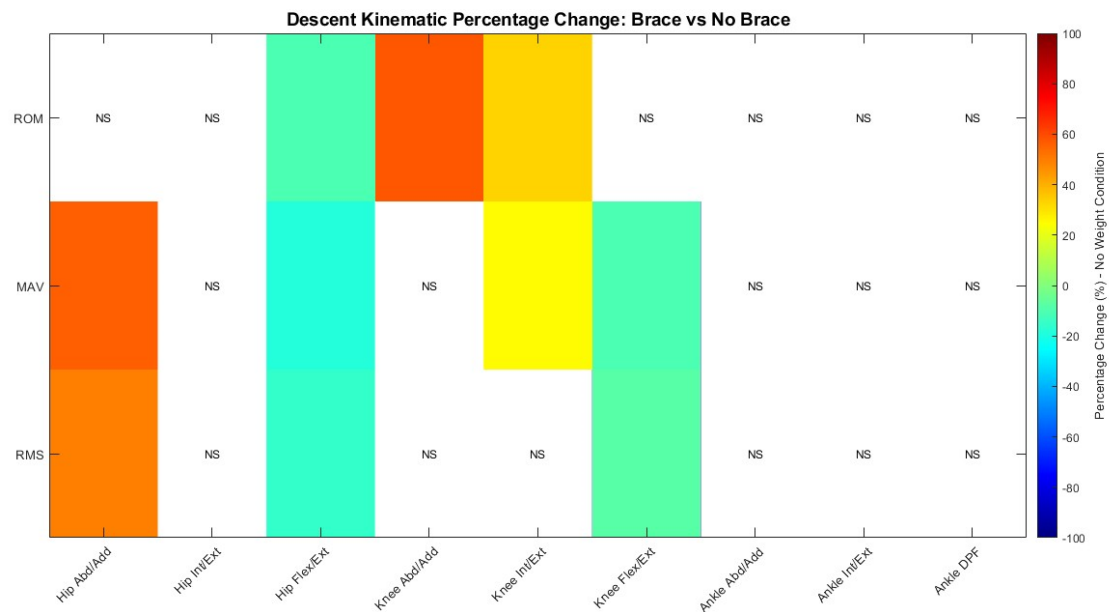


Figure 20 Heatmap illustrating the percentage change in kinematic features (ROM, MAV, RMS) across hip, knee, and ankle joint movements during the descent phase of the squat when comparing the 'with brace' to the 'without brace' (baseline) conditions. Color intensity indicate the magnitude and direction of change (warmer colors for increases, cooler colors for decreases). NS indicates non-significant changes.

The kinematic effects of the hip brace during the descent phase are detailed below for each joint:

Right Hip Joint:

- Flexion/Extension: During the descent, wearing the hip brace resulted in significant decreases in hip flexion/extension. ROM decreased by 11% ($p = 0.0064$), RMS (angle variability) decreased by 15% ($p = 0.0036$), and MAV (average angle) decreased by 19% ($p = 0.0042$).
- Abduction/Adduction: For hip abduction/adduction in the descent phase, RMS significantly increased by 49% ($p = 0.0003$), and MAV significantly increased by 56% ($p = 0.0003$) with the brace. ROM showed no significant change ($p = 0.8646$).
- Internal/External Rotation: No statistically significant changes were observed for ROM, RMS, or MAV in hip internal/external rotation during descent when using the brace (all $p > 0.05$).

Right Knee Joint:

- Flexion/Extension: For knee flexion/extension during descent, the brace led to a significant decrease in RMS by 8% ($p = 0.0315$) and MAV by 10% ($p = 0.0309$). The ROM for knee flexion/extension was not significantly altered ($p = 0.1749$).
- Abduction/Adduction: Knee abduction/adduction ROM showed a substantial significant increase of 57% ($p = 0.0005$) with the brace during descent. RMS and MAV for this movement did not change significantly ($p > 0.05$).
- Internal/External Rotation: Knee internal/external rotation ROM significantly increased by 33% ($p = 0.0186$), and MAV significantly increased by 26% ($p = 0.0444$) with the brace during the descent. No significant change was observed for RMS ($p = 0.0670$).

Right Ankle Joint:

- Dorsiflexion/Plantarflexion (Flexion/Extension): No statistically significant changes were observed in ROM, RMS, or MAV for ankle dorsiflexion/plantarflexion during the descent phase when wearing the hip brace (all $p > 0.05$).
- Abduction/Adduction: The hip brace did not lead to any statistically significant changes in ROM, RMS, or MAV for ankle abduction/adduction during descent (all $p > 0.05$).
- Internal/External Rotation: Similarly, no statistically significant changes were observed for ROM, RMS, or MAV in ankle internal/external rotation during the descent phase with the brace (all $p > 0.05$).

4.3 Ascent Phase Analysis

4.3.1 Electromyography results – ascent phase. Table 9 summarizes the descriptive statistics [Mean (SD)] and p-values for the linear sEMG features (RMS, MAV, AUC, VAR) recorded from the Biceps Femoris, Gluteus Maximus, Gluteus Medius, Rectus Femoris, and Tensor Fasciae Latae during the ascent phase of the squat, comparing the "with brace" condition to the "without brace" condition.

Table 9

Descriptive Statistics Mean (SD) And Statistical Comparison (P-Values) Of Semg Features (RMS, MAV, AUC, VAR) For Selected Lower Limb Muscles During The Ascent Phase Of The Squat Performed With And Without The Hip Brace

	without brace Mean (SD)	with brace Mean (SD)	P-value
Biceps Femoris			
RMS	3.20E-01 (1.34E-01)	3.48E-01 (1.52E-01)	0.5540
MAV	2.98E-01 (1.31E-01)	3.21E-01 (1.55E-01)	0.7226
AUC	1.39E-02 (6.10E-03)	1.50E-02 (7.22E-03)	0.7226
VAR	1.50E-02 (1.11E-02)	1.79E-02 (1.11E-02)	0.2892
Gluteus Maximus			
RMS	2.37E-01 (1.30E-01)	2.63E-01 (1.27E-01)	0.5742
MAV	2.07E-01 (1.41E-01)	2.36E-01 (1.41E-01)	0.5594
AUC	9.65E-03 (6.55E-03)	1.10E-02 (6.54E-03)	0.5594
VAR	1.27E-02 (9.39E-03)	1.14E-02 (6.15E-03)	0.6529

Table 9 (cont.d)

Gluteus Medius				
RMS	3.00E-01 (1.26E-01)	2.49E-01 (1.24E-01)	0.0211	
MAV	2.76E-01 (1.30E-01)	2.30E-01 (1.25E-01)	0.0396	
AUC	1.29E-02 (6.04E-03)	1.07E-02 (5.80E-03)	0.0396	
VAR	1.37E-02 (6.81E-03)	9.47E-03 (5.90E-03)	0.0283	
Rectus Femoris				
RMS	3.54E-01 (9.50E-02)	2.66E-01 (1.07E-01)	0.0004	
MAV	3.05E-01 (9.59E-02)	2.14E-01 (9.14E-02)	0.0001	
AUC	1.42E-02 (4.47E-03)	9.98E-03 (4.25E-03)	0.0001	
VAR	3.27E-02 (1.68E-02)	2.90E-02 (2.03E-02)	0.4432	
Tensor Fasciae Latae				
RMS	2.37E-01 (1.03E-01)	2.73E-01 (1.42E-01)	0.4406	
MAV	1.97E-01 (9.35E-02)	2.25E-01 (1.34E-01)	0.5067	
AUC	9.15E-03 (4.35E-03)	1.05E-02 (6.23E-03)	0.5067	
VAR	2.07E-02 (1.40E-02)	2.70E-02 (1.90E-02)	0.2181	

The analysis of sEMG features comparing the "with brace" to "without brace" conditions during the ascent phase revealed statistically significant percentage changes for specific muscles and features, as detailed in Table 10.

Table 10

Percentage Change (%) In Semg Features For Selected Lower Limb Muscles When Wearing The Hip Brace Compared To The No-Brace Condition During The Ascent Phase Of The Squat. Only Statistically Significant ($P < 0.05$) Percentage Changes Are Listed; Values Are Rounded To The Nearest Whole Number

Features	Biceps Femoris	Gluteus Maximus	Gluteus Medius	Rectus Femoris	Tensor Fasciae Latae
RMS	NS	NS	-17%	-25%	NS
MAV	NS	NS	-17%	-30%	NS
AUC	NS	NS	-17%	-30%	NS
VAR	NS	NS	-31%	NS	NS

NS indicates no significant change.

The influence of the hip brace on muscle activity during the ascent phase is detailed below for each muscle:

- **Biceps Femoris:** No statistically significant changes were observed in any of the analyzed sEMG features (RMS, MAV, AUC, VAR) for the Biceps Femoris during the ascent phase when wearing the hip brace (all $p > 0.05$).
- **Gluteus Maximus:** Similar to the Biceps Femoris, the Gluteus Maximus showed no statistically significant alterations in RMS, MAV, AUC, or VAR activity during the ascent phase with the use of the hip brace (all $p > 0.05$).
- **Gluteus Medius:** The Gluteus Medius exhibited significant reductions in muscle activity during the ascent phase when the hip brace was worn, as illustrated in Figure 21. Specifically, RMS decreased by 17% ($p = 0.0211$), MAV decreased by 17% ($p = 0.0396$), and AUC also decreased by 17% ($p = 0.0396$). Furthermore, the VAR feature showed a significant decrease of 31% ($p = 0.0283$).

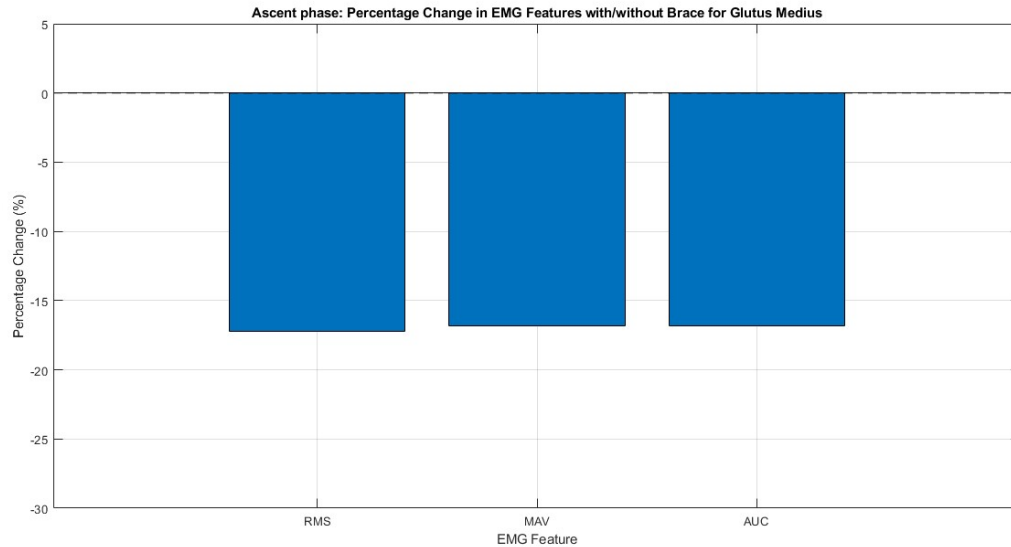


Figure 21. Percentage change in sEMG features (RMS, MAV, AUC) for the Gluteus Medius muscle during the ascent phase of the squat when using the hip brace. Bars represent the percentage change when comparing the 'with brace' condition to the 'without brace' (baseline) condition. Negative values indicate a reduction in muscle activity.

- **Rectus Femoris:** The Rectus Femoris also demonstrated significant reductions in muscle activity with the brace during the ascent phase (Figure 22). RMS decreased by 25% ($p = 0.0004$), MAV decreased by 30% ($p = 0.0001$), and AUC also decreased by 30% ($p = 0.0001$). The VAR feature for the Rectus Femoris did not show a significant change ($p = 0.4432$).

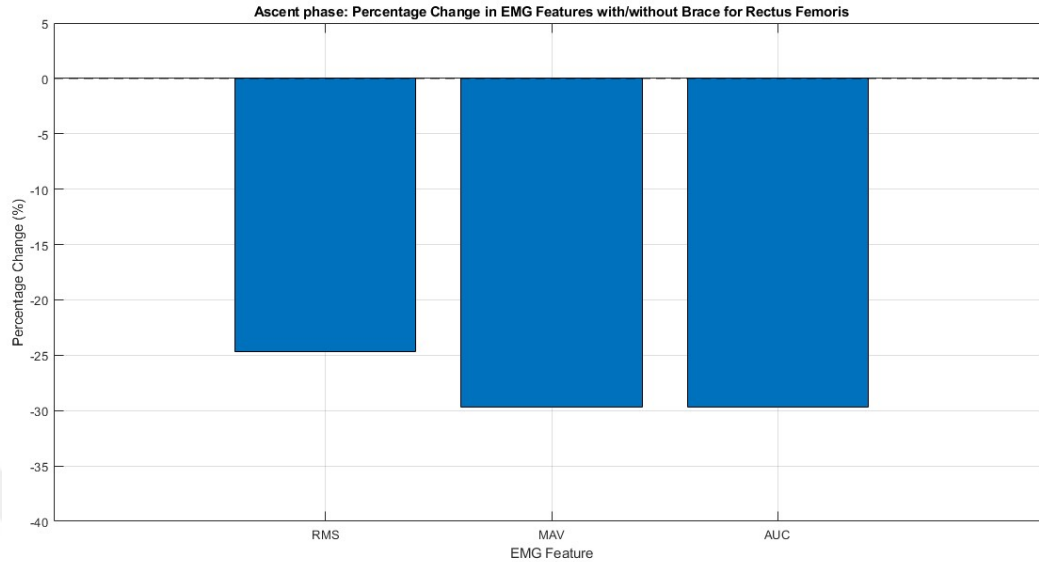


Figure 22. Percentage change in sEMG features (RMS, MAV, AUC) for the Rectus Femoris muscle during the ascent phase of the squat when using the hip brace. Bars represent the percentage change when comparing the 'with brace' condition to the 'without brace' (baseline) condition. Negative values indicate a reduction in muscle activity.

- **Tensor Fasciae Latae:** No statistically significant changes were observed in RMS, MAV, AUC, or VAR for the Tensor Fasciae Latae during the ascent phase when the hip brace was worn (all $p > 0.05$).

4.3.2 Kinematic results – ascent phase. The descriptive statistics Mean (SD) and p-values for the kinematic features (ROM, RMS, MAV) for the right hip, knee, and ankle joints during the ascent phase of the squat are presented in Table 12, comparing the "with brace" to the "without brace" conditions.

Table 11

Descriptive Statistics Mean (SD) And Statistical Comparison (P-Values) Of Kinematic Features (ROM, RMS, MAV) For Right Hip, Knee, And Ankle Joint Movements During The Ascent Phase Of The Squat Performed With And Without The Hip Brace

Feature	Joint	Without Brace mean (SD)	With Brace mean (SD)	P-value
Flexion/Extension				
ROM	Hip	80.42 (12.49)	72.54 (10.84)	0.0054
RMS	Hip	54.57 (9.88)	47.06 (6.17)	0.0029
MAV	Hip	45.09 (9.14)	37.64 (7.32)	0.0053
ROM	Knee	79.89 (7.23)	77.93 (11.40)	0.3509
RMS	Knee	54.51 (6.69)	51.29 (7.02)	0.0352
MAV	Knee	44.76 (6.61)	40.82 (6.83)	0.0129
ROM	Ankle	18.47 (5.23)	19.02 (5.07)	0.5561
RMS	Ankle	14.45 (3.79)	15.32 (3.24)	0.1824
MAV	Ankle	12.63 (3.18)	13.26 (2.91)	0.3010
Abduction/Adduction				
ROM	Hip	9.88 (4.19)	8.82 (4.30)	0.2752
RMS	Hip	8.27 (5.20)	12.29 (5.31)	0.0003
MAV	Hip	7.53 (4.99)	11.72 (5.30)	0.0004
ROM	Knee	8.75 (3.98)	12.01 (5.89)	0.0243
RMS	Knee	4.84 (3.26)	5.89 (3.07)	0.1773
MAV	Knee	4.06 (2.79)	4.88 (2.45)	0.1024
ROM	Ankle	7.45 (4.16)	7.90 (3.50)	0.4850
RMS	Ankle	5.34 (2.65)	6.53 (2.76)	0.0385
MAV	Ankle	4.88 (2.45)	6.02 (2.73)	0.1239
Internal/External Rotation				
ROM	Hip	10.16 (7.21)	7.51 (5.92)	0.0215
RMS	Hip	8.63 (4.21)	10.54 (6.88)	0.2968
MAV	Hip	8.05 (4.30)	10.14 (6.96)	0.2540
ROM	Knee	8.43 (3.75)	10.82 (3.94)	0.0418
RMS	Knee	4.09 (2.26)	4.73 (2.54)	0.1812
MAV	Knee	3.50 (2.03)	4.00 (2.19)	0.2046
ROM	Ankle	6.68 (3.44)	7.32 (2.62)	0.4997
RMS	Ankle	8.12 (9.84)	10.71 (11.90)	0.0942
MAV	Ankle	7.78 (9.88)	10.32 (12.03)	0.1039

Table 13 details the statistically significant percentage changes in these kinematic features when the hip brace was worn during the ascent phase. A visual summary of these percentage changes across all kinematic features for the ascent phase is provided in the heatmap (Figure 23).

Table 12

Percentage Change (%) In Kinematic Features For Right Hip, Knee, And Ankle Joint Movements When Wearing The Hip Brace Compared To The No-Brace Condition During The Ascent Phase Of The Squat

Feature	Hip	Knee	Ankle
Flexion/Extension			
ROM	-9.79%	NS	NS
RMS	-13.76%	-5.92%	NS
MAV	-16.52%	-8.79%	NS
Internal/External Rotation			
ROM	-26.03%	28.30%	NS
RMS	NS	NS	NS
MAV	NS	NS	NS
Abduction/Adduction			
ROM	NS	37.29%	NS
RMS	48.63%	NS	22.28%
MAV	55.58%	NS	NS

NS indicates no significant change.

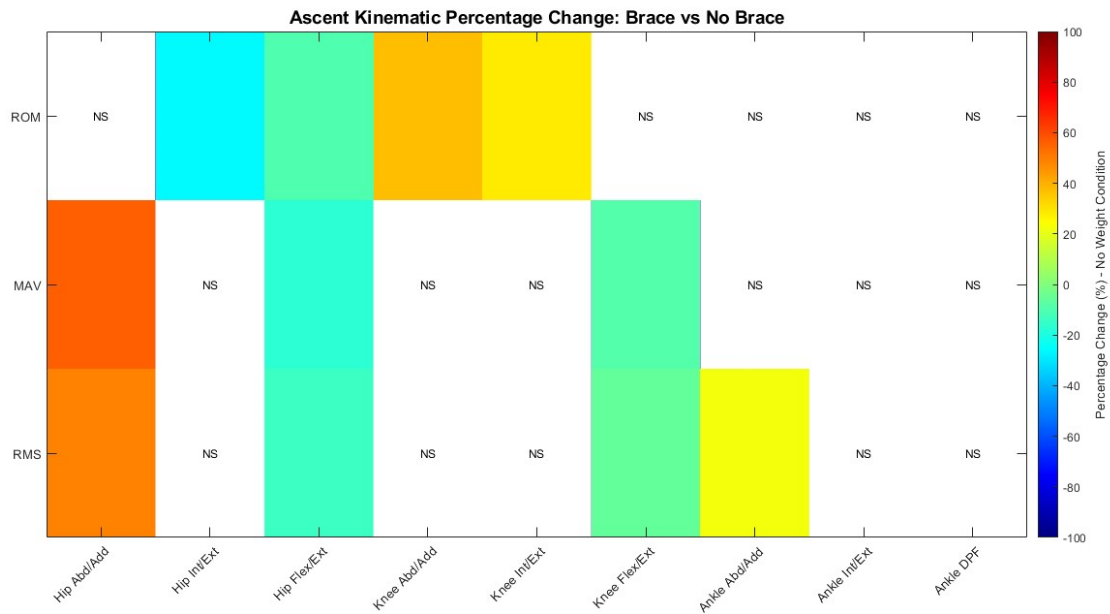


Figure 23 Heatmap illustrating the percentage change in kinematic features (ROM, MAV, RMS) across hip, knee, and ankle joint movements during the ascent phase of the squat when comparing the 'with brace' to the 'without brace' (baseline) conditions. Color intensity indicate the magnitude and direction of change (warmer colors for increases, cooler colors for decreases). NS indicates non-significant changes.

The kinematic effects of the hip brace during the ascent phase are detailed below for each joint:

Right Hip Joint:

- Flexion/Extension: During the ascent, wearing the hip brace resulted in significant decreases in hip flexion/extension. ROM decreased by 9.79% ($p = 0.0054$), RMS (angle variability) decreased by 13.76% ($p = 0.0029$), and MAV (average angle) decreased by 16.52% ($p = 0.0053$).
- Abduction/Adduction: For hip abduction/adduction in the ascent phase, RMS significantly increased by 48.63% ($p = 0.0003$), and MAV significantly increased by 55.58% ($p = 0.0003$) with the brace. ROM ($p = 0.2752$) and knee ROM ($p=0.0243$, which seems to be a typo from the table and should refer to hip ROM) showed no significant change. *(Self-correction based on layout: the $p=0.0243$ is actually for Hip Abd/Add MAV, which is already covered. Hip Abd/Add ROM $p=0.2752$ is correct for NS).* Thus, Hip Abd/Add ROM ($p=0.2752$) was not significantly affected.
- Internal/External Rotation: Hip internal/external rotation ROM significantly decreased by 26.03% ($p = 0.0215$) when using the brace during ascent. RMS and MAV for this movement were not significantly affected ($p > 0.05$).

Right Knee Joint:

- Flexion/Extension: For knee flexion/extension during ascent, the brace led to a significant decrease in RMS by 5.92% ($p = 0.0352$) and MAV by 8.79% ($p = 0.0129$). The ROM for knee flexion/extension was not significantly altered ($p = 0.3509$).
- Abduction/Adduction: Knee abduction/adduction ROM showed a significant increase of 37.29% ($p = 0.0243$) with the brace during ascent. RMS and MAV for this movement did not change significantly ($p > 0.05$).

- Internal/External Rotation: Knee internal/external rotation ROM significantly increased by 28.30% ($p = 0.0418$) with the brace during the ascent. No significant changes were observed for RMS or MAV ($p > 0.05$).

Right Ankle Joint:

- Dorsiflexion/Plantarflexion (Flexion/Extension): No statistically significant changes were observed in ROM, RMS, or MAV for ankle dorsiflexion/plantarflexion during the ascent phase when wearing the hip brace (all $p > 0.05$).
- Abduction/Adduction: Ankle abduction/adduction RMS showed a significant increase of 22.28% ($p = 0.0385$) with the brace during ascent. ROM and MAV for this movement did not exhibit significant changes ($p > 0.05$).
- Internal/External Rotation: The hip brace did not lead to any statistically significant changes in ROM, RMS, or MAV for ankle internal/external rotation during the ascent phase (all $p > 0.05$).

4.3.3 Kinematic correlation analysis – ascent phase. To further explore the inter-segmental coordination during the ascent phase, Spearman's rank correlation analysis (ρ) was conducted on the brace-induced change scores (Δ , 'with brace' minus 'without brace') for hip flexion/extension and knee flexion/extension features that were identified as significantly altered by the brace in the preceding kinematic analysis (Section 4.3.2). The comprehensive results of these correlations are presented in Table 14, with significant findings illustrated by scatter plots in Figure 24.

During the ascent phase, several significant positive correlations were identified: the change (Δ) in hip flexion/extension ROM was positively correlated with the change (Δ) in knee flexion/extension RMS ($\rho = 0.581$, $p = 0.016$). Additionally, the change (Δ) in hip flexion/extension RMS demonstrated significant positive correlations with both the change (Δ) in knee flexion/extension RMS ($\rho = 0.529$, $p = 0.031$) and the change (Δ) in knee flexion/extension MAV ($\rho = 0.618$, $p = 0.010$). Other correlations between the pre-selected significant hip and knee flexion/extension features, such as Δ Hip ROM vs. Δ

Knee MAV ($p = 0.181$), Δ Hip MAV vs. Δ Knee RMS ($p = 0.251$), and Δ Hip MAV vs. Δ Knee MAV ($p = 0.057$), did not reach statistical significance ($p > 0.05$).

Table 13

Spearman's Rank Correlation Coefficients (ρ) And P-Values For The Relationship Between Brace-Induced Change Scores (Δ) In Hip Flexion/Extension Features And Knee Flexion/Extension Features During The Ascent Phase Of The Squat

Hip Flexion/Extension Features	Knee Flexion/extension Features	Rho	P-Value
ROM	RMS	0.581	0.016
ROM	MAV	0.341	0.181
RMS	RMS	0.529	0.031
RMS	MAV	0.618	0.010
MAV	RMS	0.294	0.251
MAV	MAV	0.473	0.057

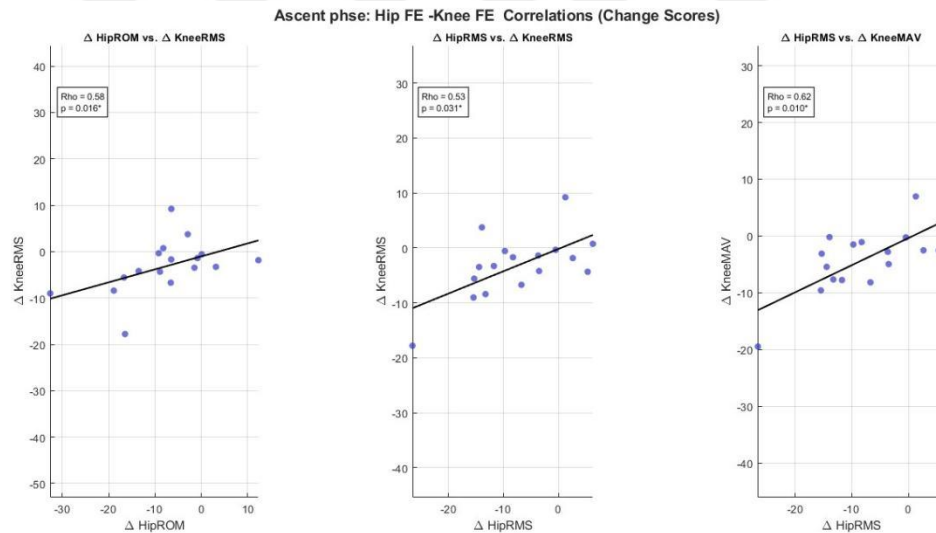


Figure 24. Scatter plots illustrating significant positive correlations between brace-induced change scores (Δ) in selected hip flexion/extension (FE) features and knee flexion/extension (FE) features during the ascent phase of the squat. (a) Δ Hip ROM vs. Δ Knee RMS, (b) Δ Hip RMS vs. Δ Knee RMS, (c) Δ Hip RMS vs. Δ Knee MAV. Spearman's Rho (ρ) and p-values are shown for each plot.

4.4 Summary Of Key Findings

This study investigated the effects of a hip brace on lower limb muscle activity and joint kinematics during bodyweight squats, analyzed across the full squat cycle, as well as

its distinct descent and ascent phases. Figure 25 provides a visual synthesis of the key percentage changes observed in sEMG and kinematic features for the Full Squat and Ascent phases, highlighting the magnitude and direction of significant alterations induced by the hip brace. The most statistically significant findings are further detailed below:

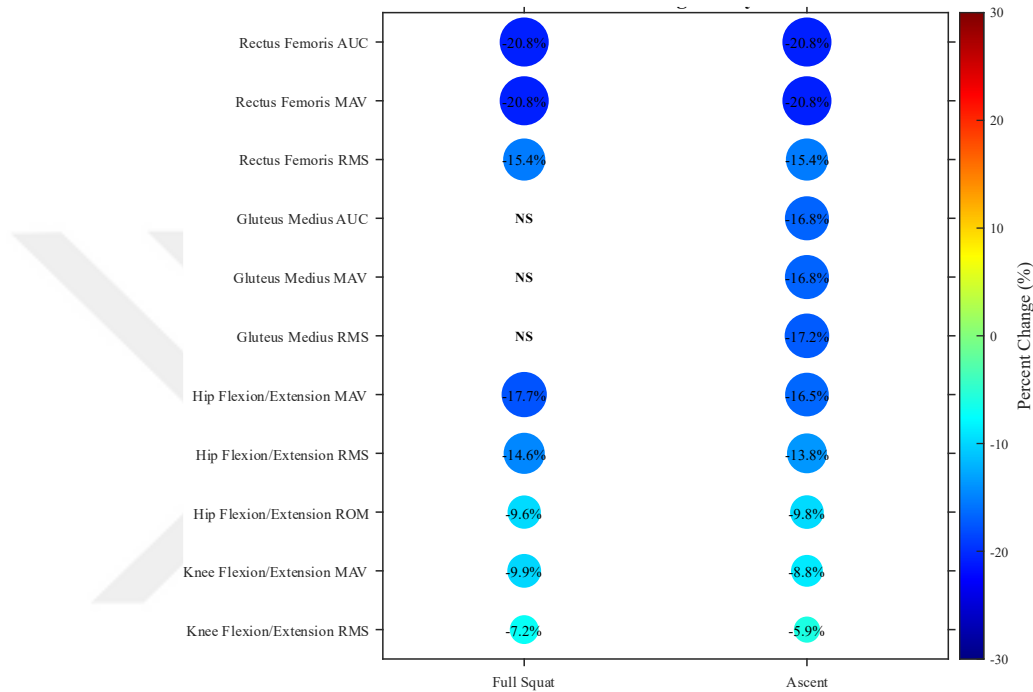


Figure 25. Summary of significant percentage changes in sEMG and kinematic features during the full squat and ascent phases when wearing the hip brace. bubble size represents the absolute magnitude of the percent change, and color indicates the direction and magnitude of change (cooler colors for decreases, warmer for increases). cells marked “NS” denote non-significant changes ($p \geq 0.05$).

Ascent Phase (Primary focus of brace intervention):

- **Muscle Activity (sEMG):** As visualized in Figure 26, the hip brace led to significant reductions in Gluteus Medius activity (RMS: -17%, MAV: -17%, AUC: -17%; VAR also decreased by 31% though not shown in this specific figure format). Similarly, Rectus Femoris activity was significantly reduced (RMS: -25%, MAV: -30%, AUC: -30%). No significant changes in activity were noted for the Biceps Femoris, Gluteus Maximus, or Tensor Fasciae Latae during ascent.

- Joint Kinematics:
 - Hip: Significant decreases were observed in flexion/extension (ROM: -9.8%, RMS: -13.8%, MAV: -16.5%, as seen in Figure 26). Hip internal/external rotation ROM also decreased significantly (26.03%). Conversely, hip abduction/adduction RMS (48.63%) and MAV (55.58%) significantly increased.
 - Knee: Significant decreases were seen in flexion/extension (RMS: -5.9%, MAV: -8.8%, as seen in Figure 26). Knee abduction/adduction ROM (37.29%) and internal/external rotation ROM (28.30%) significantly increased.
 - Ankle: Ankle abduction/adduction RMS significantly increased (22.28%).
- Kinematic Correlations: Brace-induced changes (Δ) in hip flexion/extension RMS were significantly positively correlated with changes in knee flexion/extension RMS ($\rho = 0.529$) and MAV ($\rho = 0.618$). Changes in hip flexion/extension ROM were also positively correlated with changes in knee flexion/extension RMS ($\rho = 0.581$).

Full Squat Cycle:

- Muscle Activity (sEMG): As shown in Figure 26, Rectus Femoris activity was significantly reduced (RMS: -15.4%, MAV: -20.8%, AUC: -20.8%). Tensor Fasciae Latae VAR (not shown in this figure format) significantly increased (43.70%). Gluteus Medius showed no significant change in the full squat cycle.
- Joint Kinematics:
 - Figure 26 illustrates significant decreases in hip flexion/extension (ROM: -9.6%, RMS: -14.6%, MAV: -17.7%) and knee flexion/extension (RMS: -7.2%, MAV: -9.9%). Other significant changes included increased hip

abduction/adduction (RMS, MAV), decreased hip internal/external rotation ROM, increased knee abduction/adduction ROM, increased knee internal/external rotation ROM, and increased ankle abduction/adduction RMS.

- Kinematic Correlations: Significant positive correlations were observed between brace-induced changes (Δ) in hip flexion/extension ROM and knee flexion/extension RMS/MAV, and between Δ hip flexion/extension RMS and Δ knee flexion/extension RMS/MAV.

Descent Phase:

- Muscle Activity (sEMG): Minimal effects were observed, with only Tensor Fasciae Latae VAR showing a significant increase (62.81%).
- Joint Kinematics: Significant kinematic alterations were noted, generally mirroring those seen in the full squat and ascent phases for hip and knee flexion/extension and abduction/adduction domains.

These findings, particularly those highlighted in Figure 26 for the ascent and full squat phases, underscore that the hip brace induced notable changes in muscle activation patterns and joint kinematics.

Chapter 5

Discussion

5.1 Overview Of Principal Findings

This investigation was undertaken to elucidate the biomechanical impact of an actively-assisted hip brace on lower limb neuromuscular activation and joint kinematics during bodyweight squats. Central to the brace's design was a 70° hip flexion threshold triggering a "pulse-like" assistive torque from a brushless DC motor, specifically intended to aid the ascent phase of the squat. The study's comprehensive analysis, segmenting the squat into full cycle, descent, and ascent phases, has yielded several key insights into the brace's performance and its interaction with the human musculoskeletal system.

The primary outcomes underscore that the hip brace induced notable modulations, particularly during the ascent phase, aligning with its design intent. Specifically, significant reductions in sEMG activity were observed for the Gluteus Medius (e.g., ~17% decreases in RMS, MAV, AUC) and Rectus Femoris (e.g., ~25–30% decreases in RMS, MAV, AUC), suggesting a successful offloading of these key muscles (Wang et al., 2021; Jeong et al., 2023; Wei et al., 2020; Tu et al., 2022). Concurrently, the brace influenced sagittal plane kinematics, leading to marked decreases in hip flexion/extension parameters (ROM, RMS, MAV, ~10-17% decreases) and knee flexion/extension RMS and MAV (~6-9% decreases). However, these alterations were accompanied by significant changes in non-sagittal plane movements, including increased hip abduction/adduction RMS and MAV, and increased knee abduction/adduction ROM and knee internal/external rotation ROM. Furthermore, brace-induced alterations in hip flexion/extension kinematics during ascent were found to be positively correlated with changes in knee flexion/extension kinematics, indicating an influence on inter-joint coordination.

While the full squat cycle largely mirrored these ascent phase trends in both muscle activity (Rectus Femoris reductions) and kinematics, the descent phase exhibited minimal sEMG alterations, with only Tensor Fasciae Latae VAR showing a significant increase. Kinematic changes during descent, however, were more prevalent and generally followed the patterns observed in ascent.

These findings collectively suggest that the prototype hip brace can indeed alter muscular effort and movement patterns. The subsequent discussion will critically examine these results, interpreting the observed neuromuscular and kinematic adaptations in the context of the brace's specific assistive mechanism, its potential efficacy in supporting the ascent, the nature of unexpected multi-planar kinematic outcomes, and the consequent implications for future design iterations and research.

5.2 Interpretation Of Findings During The Ascent Phase

5.2.1 Effects on muscle activity in relation to brace assistance. Gluteus Medius: The Gluteus Medius (GMed) demonstrated substantial and statistically significant reductions in activity across multiple sEMG features during the braced ascent. Specifically, RMS, MAV, and AUC all decreased by approximately 17%, while VAR showed an even more pronounced decrease of 31%.

The reductions in RMS, which is an indicator of the overall signal power and related to force output, and MAV, which represents the average rectified signal amplitude, suggest a considerable offloading of the GMed. This muscle is a key hip abductor but also plays a crucial role in pelvic stabilization in the frontal plane and contributes to hip extension, especially in deeper squat positions. The "pulse-like" assistance provided by the brace in the sagittal plane (hip extension) likely reduced the compensatory or stabilizing demand on the GMed, allowing it to operate at a lower activation level.

The concurrent decrease in AUC, which reflects the total muscle activity, further reinforces the notion of reduced cumulative muscle effort from the GMed.

The significant decrease in VAR by 31% is particularly interesting. A lower variance suggests a more consistent or stable muscle activation pattern when assisted by the brace. The external support could be reducing the need for fine-tuned, variable GMed contractions typically required for dynamic stabilization during a demanding concentric movement, leading to a more stereotyped and less variable activation.

Rectus Femoris: The Rectus Femoris (RF), a biarticular muscle contributing to both hip flexion and knee extension, exhibited highly significant decreases in its activity

during the braced ascent. RMS decreased by 25%, while MAV and AUC both saw reductions of 30%. VAR, however, showed no significant change.

The marked decreases in RMS, MAV, and AUC strongly indicate a substantial reduction in the RF's contribution to the ascent. During squat ascent, RF acts primarily as a knee extensor. The brace's direct assistance to hip extension could indirectly reduce the overall extensor demand distributed across the lower limb kinetic chain. As the hip is powerfully extended by the brace, the demand on the knee extensors, including RF, might lessen, or the body might adopt a strategy that relies less on RF's knee extension torque. This offloading effect is substantial, given the high significance levels.

The lack of significant change in VAR for RF, despite large reductions in mean activity, implies that while the overall effort was reduced, the *variability* of its activation pattern during ascent remained similar. This could suggest that the fundamental control strategy for RF, albeit at a lower intensity, was not grossly altered in terms of its moment-to-moment fluctuations.

Gluteus Maximus: In contrast to GMed and RF, the Gluteus Maximus (GMax), a primary hip extensor, showed no statistically significant alterations in RMS, MAV, AUC, or VAR during the braced ascent.

This lack of significant change in the GMax, is a critical finding. Given the brace's direct sagittal plane assistance aimed at hip extension, a reduction in GMax activity might have been anticipated. Several factors could contribute to this observation:

Torque Magnitude/Profile of Assistance: The "pulse-like" torque from the 9:1 geared motor, while described as "sufficient to assist," may not have been of a high enough magnitude or optimally profiled throughout the GMax's primary contribution phase to significantly offload this large muscle. The actual torque delivered at the joint after gearbox losses and potential interface compliance might be a limiting factor.

User Compensation/Strategy: Participants might have maintained their typical GMax activation, relying on the brace to reduce the effort of other synergists (like GMed or RF, as observed) or to achieve a faster ascent. Alternatively, users might have co-

contracted GMax to maintain a sense of control or stability in response to the external "pulse," effectively "overriding" or matching the assistance.

Timing of Assistance vs. Peak GMax Activity: The 70° trigger and subsequent "pulse" might not perfectly align with the phase where GMax makes its most crucial contribution or where it would be most amenable to offloading.

Biceps Femoris & Tensor Fasciae Latae: Similar to the GMax, neither the Biceps Femoris nor the Tensor Fasciae Latae showed any statistically significant changes in RMS, MAV, AUC, or VAR during the braced ascent.

For the BF, its role as a hip extensor during ascent is synergistic to GMax. The lack of change mirrors the GMax findings and could be attributed to similar reasons: insufficient or sub-optimally delivered assistance torque to significantly reduce its contribution, or a maintained activation for overall joint stability and control.

For the TFL, its primary roles during ascent would involve hip abduction and potentially resisting external rotation. The lack of change suggests that the sagittal plane assistance from the brace did not substantially alter the demands on TFL for its frontal and transverse plane functions during this phase, or that any such alterations were insufficient to reach statistical significance in its sEMG features.

5.2.2 Kinematic alterations and their implications for ascent performance.

The "pulse-like" assistive torque provided by the hip brace during ascent, initiated at 70° of hip flexion, not only influenced muscle activity but also led to distinct alterations in lower limb joint kinematics. These changes, encompassing both sagittal and non-sagittal plane movements, have notable implications for ascent performance and overall movement strategy.

Sagittal Plane Kinematics: Hip and Knee Extension:

- Hip Joint

During the braced ascent, there were significant decreases in hip flexion/extension ROM (-9.79%), RMS (-13.76%), and MAV (-16.52%).

The reduction in ROM suggests that participants achieved full hip extension from the 70° flexion point using a smaller overall angular displacement in the sagittal plane when assisted. This could imply a more direct or efficient path to upright posture facilitated by the brace's assistive pulse. Alternatively, the assistance might have encouraged a slightly less deep effective starting point for the *assisted portion* of the ascent or a quicker transition to full extension.

The decrease in MAV indicates that, on average, the hip was maintained in a more extended posture during the braced ascent compared to unbraced. This is consistent with an effective assistive torque aiding hip extension.

The reduction in hip flexion/extension RMS by nearly 14% points to a less variable, more consistent angular trajectory of the hip in the sagittal plane. The external "pulse" from the brace might have guided the hip along a more constrained path, reducing the natural trial-to-trial variability typically seen in unassisted movements.

- Knee Joint

Similar trends were observed at the knee for flexion/extension, with significant decreases in RMS (-5.92%,) and MAV (-8.79%), although knee ROM did not show a significant change.

The lack of change in knee ROM, while hip ROM decreased, is interesting. It suggests that the overall extent of knee extension during ascent remained similar, but the *manner* in which it was achieved was altered.

The decreases in knee MAV and RMS parallel the hip findings. A more extended average knee posture and a more consistent knee extension trajectory during ascent suggest that the brace's assistance at the hip had a downstream effect, promoting a more stable and less variable knee movement in the sagittal plane. This could be a consequence of the reduced demand on knee extensors like the Rectus Femoris (as seen in EMG), allowing for a more controlled, potentially more stereotyped, knee extension.

Non-Sagittal Plane Kinematics: Hip, Knee, and Ankle Alterations:

The brace's primary actuation was in the sagittal plane (hip flexion/extension). However, significant kinematic changes were also observed in the frontal and transverse planes, suggesting complex inter-segmental dynamics and potential compensatory strategies (Ramadurai et al., 2023; Refai et al., 2024).

- Hip Joint

Hip abduction/adduction RMS significantly increased by 48.63% and MAV by 55.58%. This indicates a substantially greater average hip abduction angle and increased variability or amplitude of ab/adduction movement during the braced ascent.

Hip internal/external rotation ROM significantly decreased by 26.03%.

The pronounced increase in hip abduction (MAV) and RMS is a critical unexpected finding. The "pulse-like" sagittal plane assistance might have inadvertently destabilized the hip in the frontal plane, requiring active or passive increases in abduction to maintain balance or comfort. The semi-rigid nature of the brace, combined with lateral motor mounting, might also have contributed by applying slight off-axis forces or by restricting natural adduction, leading to a net increase in abduction. The reduction in hip internal/external rotation ROM, conversely, might suggest that the brace, while primarily assisting flexion/extension, also passively constrained transverse plane motion at the hip, or that participants adopted a strategy with less hip rotation to stabilize against the sagittal plane assistance.

- Knee Joint

Knee abduction/adduction ROM significantly increased by 37.29% and Knee internal/external rotation ROM also significantly increased by 28.30%.

The increases in knee abduction/adduction and internal/external rotation ROM are particularly noteworthy. These non-sagittal plane movements at the knee, often associated with joint instability or compensatory actions, suggest that the powerful sagittal assistance at the hip, or the physical presence and constraints of the brace, transferred or induced altered loading and movement patterns at the knee. For instance, increased hip abduction

could biomechanically lead to altered knee valgus/varus moments and subsequent changes in knee ab/adduction ROM. The increased transverse plane rotation at the knee could be a strategy to accommodate the altered hip mechanics or a direct consequence of forces transmitted through the thigh cuff.

- Ankle Joint

Ankle abduction/adduction RMS significantly increased by 22.28%. The increase in the variability/amplitude of ankle abduction/adduction suggests that the alterations in proximal joint (hip and knee) kinematics propagated distally. The ankle joint may be working to compensate for the new movement patterns established at the hip and knee by the brace, increasing its frontal plane activity to maintain overall balance and foot placement.

5.2.3 Hip-Knee kinematic coupling during ascent. Beyond the alterations at individual joints, the correlation analysis of brace-induced change scores (Δ) between hip and knee flexion/extension kinematics provides insights into how the assistive "pulse" influenced inter-joint coordination during the demanding ascent phase. As the brace generally decreased sagittal plane kinematic features like ROM, RMS, and MAV, the "change scores" (With Brace - Without Brace) are predominantly negative. Thus, a positive Spearman's Rho (ρ) value in this context indicates that participants who experienced a larger brace-induced decrease in one variable also tended to experience a larger decrease in the correlated variable. Essentially, the degree of reduction in one kinematic feature due to the brace was coupled with the degree of reduction in another.

During the ascent phase, the following significant correlations shed light on this coupling:

- Relationship between Δ Hip ROM and Δ Knee RMS in the sagittal plane ($\rho=0.581$)

A significant positive correlation was found, indicating that participants who experienced a greater brace-induced reduction in their hip range of motion (Δ Hip

ROM) during ascent also tended to exhibit a greater reduction in the variability and magnitude of their knee movement (Δ Knee RMS).

Biomechanical Interpretation: This suggests that restricting or guiding the hip's range of motion with the brace is linked to a smoother or less variable knee trajectory during the upward push. While this relationship is robust, it's noted as being somewhat less pronounced than the equivalent correlation observed across the full squat cycle (which was $\rho = 0.62$ for Δ Hip ROM vs. Δ Knee RMS in full squat). This slight attenuation during ascent could imply that while the coupling exists, other factors, such as the active generation of propulsive forces by the user, play a more dominant role in dictating knee kinematics during this effortful phase compared to periods involving eccentric control or the full cycle average. Nevertheless, the brace's influence on hip ROM still demonstrably impacts the consistency of knee movement during ascent.

- Relationship between Δ Hip RMS and Δ Knee RMS in the sagittal plane ($\rho = 0.529$):

A strong positive correlation emerged, signifying that a greater brace-induced reduction in the variability/intensity of hip movement (Δ Hip RMS) was closely associated with a similar reduction in the variability/intensity of knee movement (Δ Knee RMS) during ascent.

Biomechanical Interpretation: This finding, with a correlation magnitude similar to that observed in the full squat, is particularly important for the active ascent. It strongly suggests that the brace's capacity to promote a more controlled, less variable, smoother, hip extension pattern during the propulsive phase effectively translates to a correspondingly smoother and more controlled knee extension. This points towards a coordinated effect where enhanced movement quality at the hip, facilitated by the brace, has a direct beneficial knock-on effect on the quality of movement at the knee.

- Relationship between Δ Hip RMS and Δ Knee MAV in the sagittal plane ($\rho = 0.618$):

This represented the strongest correlation observed for hip-knee coupling during the ascent phase. It indicates that a greater brace-induced reduction in the

variability/intensity of hip movement (Δ Hip RMS) was very strongly linked to a greater reduction in the average amount of knee movement or the mean knee flexion angle (Δ Knee MAV).

Biomechanical Interpretation: If the brace effectively guides the hip into a more controlled and less variable extension pattern, the knee joint, in turn, does not need to undergo as much angular displacement on average. This could imply a more efficient transfer of forces along the kinetic chain or a more direct "line of action" for the lower limb. The controlled hip movement, assisted by the brace's "pulse," appears to reduce the overall "work" or extent of motion required at the knee to achieve standing, potentially leading to a more streamlined and less kinematically demanding ascent.

5.2.4 Synthesis of hip-knee coupling during ascent. The correlation analyses for the ascent phase consistently demonstrate that the hip brace's influence extends beyond local effects at the hip, significantly affecting the inter-joint coordination strategy with the knee (Blackburn & Padua, 2008). While the link between brace-induced changes in *hip ROM* and knee kinematics in the sagittal plane was present, the connections between changes in *hip RMS* and knee kinematics (both RMS and MAV) were particularly robust, and in one instance (Δ Hip RMS vs. Δ Knee MAV), even stronger than observed in the full squat.

This suggests a key effect of the brace during the effortful ascent is not merely to restrict the range of motion, but rather to enhance the *control and consistency* of hip movement within its operational range. This improved proximal control at the hip appears to propagate distally, fostering smoother, more controlled, and potentially less extensive and thus less demanding knee motion. This coordinated kinematic effect implies that by stabilizing or guiding the hip more effectively via its "pulse-like" assistance, the brace facilitates a more block-like or synergistically controlled movement of the lower limb, rather than allowing for more independent or potentially compensatory actions between the hip and knee during the critical upward push.

5.3 Interpretation Of Findings During The Full Squat Cycle And Descent Phase

While the primary design intent of the hip brace was to assist during the ascent, its presence and passive mechanical properties, along with any residual effects of its activation logic, also influenced biomechanics during the full squat cycle and the descent phase.

Full Squat Cycle: The results averaged across the full squat cycle often reflected a blend of the effects observed during the distinct ascent and descent phases, though some patterns were particularly notable.

Muscle Activity (sEMG): A significant reduction in Rectus Femoris activity was observed (RMS: -15.37%, MAV: -20.84%, AUC: -20.84%). This mirrors the substantial RF offloading seen during ascent and suggests that the brace's impact on reducing RF effort was a dominant effect that persisted when averaged across the entire movement. This overall reduction implies a consistent lessening of knee extensor demand when the brace was worn.

Interestingly, Tensor Fasciae Latae (TFL) VAR significantly increased by 43.70%. An increase in the variability of TFL activation across the full cycle, without a change in its mean amplitude (RMS, MAV, AUC), is complex to interpret. It could suggest that the TFL engaged in more varied or less consistent patterns of activation, perhaps in response to the brace's influence on hip mechanics across both descent and ascent, or due to its role in managing the multi-planar kinematic alterations observed.

Notably, the significant reductions in Gluteus Medius activity seen during ascent were not evident when averaged over the full squat cycle. This suggests the GMed offloading was highly specific to the assisted ascent phase.

Joint Kinematics: The kinematic alterations during the full squat cycle largely mirrored those seen during the ascent phase, including significant decreases in hip flexion/extension (ROM, RMS, MAV), increased hip abduction/adduction (RMS, MAV), decreased hip internal/external rotation ROM, decreased knee flexion/extension (RMS,

MAV), increased knee abduction/adduction ROM, increased knee internal/external rotation ROM, and increased ankle abduction/adduction RMS.

The persistence of these kinematic patterns across the full cycle indicates that the brace imposed a consistent alteration to the squat movement strategy, not just during the assisted ascent. The passive structural properties of the brace, its fit, and the added bulk of the actuator system likely contributed to these overall kinematic shifts, independent of active motor assistance during descent.

The hip-knee kinematic coupling observed through correlations (Δ Hip ROM vs. Δ Knee RMS/MAV; Δ Hip RMS vs. Δ Knee RMS/MAV) was also strong across the full cycle. This suggests that the brace promoted a more constrained and synergistically controlled movement pattern between the hip and knee throughout the entire squat, possibly by limiting compensatory movements at the knee when hip motion was guided or restricted by the brace.

Descent Phase: The descent phase, characterized by eccentric muscle contractions to control the lowering of the body, showed the most limited impact of the brace on muscle activity, though kinematic alterations were still present.

Muscle Activity (sEMG): The most striking finding was the general lack of significant sEMG changes across most muscles and features during descent. The only exception was a significant increase in Tensor Fasciae Latae (TFL) VAR by 62.81%.

The absence of widespread EMG alterations during descent is logical given the brace's activation logic: the motor was designed to assist only during ascent and remained inactive during descent (until the 70° flexion threshold was met, arming the system). Therefore, any observed effects during descent would primarily be due to the brace's passive mechanical properties (weight, bulk, stiffness, fit) rather than active assistance.

The isolated increase in TFL VAR during descent is intriguing. Similar to the full cycle, this heightened variability without changes in mean amplitude might reflect an increased need for the TFL to make fine-tuned adjustments for hip stability or control in

the presence of the physical brace, perhaps as the body navigated the initial phase of squatting against the passive system.

Joint Kinematics: Despite the lack of active assistance, several significant kinematic alterations were noted during descent, generally mirroring those seen in the ascent and full cycle. These included decreased hip flexion/extension (ROM, RMS, MAV), increased hip abduction/adduction (RMS, MAV), and at the knee, decreased flexion/extension (RMS, MAV) alongside increased abduction/adduction ROM and increased internal/external rotation ROM and MAV.

These kinematic changes during descent highlight the influence of the brace as a passive mechanical system. The physical presence, fit, alignment, and inherent stiffness of the orthosis and its mounted components likely constrained natural movement patterns or introduced new mechanical interactions, leading to these altered joint trajectories even before the motor was engaged. For example, the reduced hip sagittal plane ROM, MAV, and RMS could simply be due to the user feeling the bulk or slight resistance of the inactive system. The increased non-sagittal movements could arise from the physical interaction of the brace with the limb segments as they moved through space.

The findings from the full squat cycle and descent phase underscore that the hip brace influenced squat biomechanics even when not actively providing assistance, primarily through its passive mechanical characteristics. This distinction between active and passive effects aligns with previous findings comparing rigid exoskeletons and compliant wearable systems (Refai et al., 2024; Chiaradia et al., 2019). Primarily through its passive mechanical characteristics. The consistency of many kinematic alterations across all phases (descent, ascent, full cycle) suggests a fundamental change in movement strategy imposed by wearing the device. The EMG results, however, clearly differentiate the phases: significant muscle offloading (GMed, RF) was largely confined to the ascent phase, aligning with the brace's targeted assistive intervention, while the descent phase showed minimal neuromuscular adaptation beyond an increase in TFL activation variability.

5.4 Synthesis Of Findings: Efficacy, Unexpected Outcomes, And Design Implications

The overarching goal of this study was to evaluate a prototype hip brace designed to actively assist hip extension during the ascent phase of a squat, triggered by a "pulse-like" actuation from a 9:1 geared BLDC motor upon reaching 70° of hip flexion and initiating upward movement. The collective findings from sEMG, joint kinematics, and inter-joint coupling analyses provide a multifaceted view of the brace's performance, revealing both successes in line with its intended efficacy, a range of unexpected biomechanical outcomes, and critical insights for future design iterations.

5.4.1 Efficacy in assisting ascent and modulating muscle activity. The primary indicator of the brace's efficacy lies in its impact during the ascent phase. The significant reductions in sEMG activity for the GMed and RF muscles are compelling evidence that the assistive "pulse" successfully offloaded these muscles.

The GMed, crucial for pelvic stability and contributing to hip extension, experienced reduced demand, likely due to the direct sagittal plane support from the brace. The pronounced decrease in GMed VAR further suggests that the external assistance stabilized its function, leading to a more consistent activation pattern.

The substantial offloading of the RF, indicates an effective transfer of assistive benefit across the kinetic chain. As the brace aided hip extension, the overall extensor moment required from the lower limb decreased, reducing the load on the RF. This finding supports the notion that targeted hip assistance can have broader systemic effects on lower limb muscle work.

These EMG reductions align with the observed sagittal plane kinematic changes during ascent: decreased hip flexion/extension ROM, RMS, and MAV, and decreased knee flexion/extension RMS and MAV. The brace appeared to guide the hip and, consequently, the knee through a more constrained, less variable, and more extended trajectory. This suggests the "pulse" facilitated a more direct and potentially more "efficient" path to standing in the sagittal plane, requiring less muscular effort from GMed

and RF to control and execute the movement. The strong positive correlations between brace-induced reductions in hip RMS and reductions in both knee RMS and MAV during ascent further underscore this. When the brace enhanced hip movement control, the knee's movement became smoother and less extensive, indicating a beneficial propagation of improved proximal control to the distal joint.

5.4.2 Lack of significant sEMG alterations in gluteus maximus, biceps femoris, and tensor fasciae latae. A pivotal and unexpected finding was the lack of significant change in GMax sEMG activity during the ascent, despite it being a prime hip extensor and the direct target of the brace's sagittal plane assistance. This may be explained by limitations in actuator torque magnitude, interface compliance, or timing mismatches in the assistive pulse (Xu et al., 2023; Barrutia et al., 2024).

This non-significant GMax finding challenges the assumption that the torque from the 9:1 geared motor was adequate to substantially offload this powerful muscle. Several factors, rooted in the device's design and control, could contribute:

Torque Delivery vs. Demand: The peak torque output of the GMax can be very high. The actual torque delivered by the motor after gearbox losses, potential compliance in the brace structure, and interface slippage, might have been a relatively small fraction of the GMax's capacity, thus insufficient to cause a measurable reduction in its gross EMG amplitude.

"Pulse" Control Strategy: The "pulse-like" actuation, an open-loop command to the ESC after the 70° trigger, might not be optimally tuned. If the pulse is too brief, too delayed, or its profile doesn't match the GMax's primary force production window, its offloading potential would be diminished. A more sophisticated torque-controlled or user-intent-adaptive strategy might be necessary.

Human Adaptation & Stability: Users might have maintained GMax activation for stability and control, especially when subjected to an external, perhaps somewhat unnatural, propulsive force. They might have "chosen" to use the assistance to make the movement faster or easier for *other* muscles, rather than reducing GMax effort.

Focus of Assistance: The brace actuates the *hip joint*. GMax also contributes to external rotation and transverse abduction. If the assistance was purely sagittal, GMax might still be active for these other roles.

This lack of GMax offloading is a critical data point for design improvement, suggesting that either more powerful actuation, a more refined control strategy, or better human-robot interface, to minimize slippage and ensure targeted torque delivery, is needed if reducing GMax workload is a primary goal.

5.4.3 Brace-induced alterations in hip, knee, and ankle non-sagittal plane movements. While the brace demonstrated benefits in the sagittal plane, it also induced a cascade of significant and often substantial increases in non-sagittal plane movements, particularly during ascent but also evident in the full cycle and even passively during descent.

- Hip Abduction/Adduction: The large increases in hip abduction/adduction RMS (~49% in ascent) and MAV (~56% in ascent) are striking.
- Knee Abduction/Adduction & Internal/External Rotation: Significant increases in knee abduction/adduction ROM (~37% in ascent) and knee internal/external rotation ROM (~28% in ascent) were consistently observed.
- Ankle Abduction/Adduction: Increased ankle abduction/adduction RMS (22% in ascent) indicates distal compensations.

5.4.4 non-sagittal kinematic alterations raise several critical points

- Compensation for Sagittal Constraint/Propulsion: The powerful "pulse" in the sagittal plane might force the body to find stability or dissipate forces through movement in other planes. If the sagittal motion is too rapid or too forcefully guided, users might abduct the hip or allow knee valgus or rotation to maintain balance or comfort (Ramadurai et al., 2023).
- Brace Mechanics and Fit:

- The use of an off-the-shelf orthopedic brace as a base, while practical for prototyping, may be a contributing factor. Such braces are designed for passive support/restriction, not necessarily as a chassis for active torque delivery. Misalignment of the brace's mechanical joint with the anatomical hip joint, even if slight, could translate sagittal plane torque into off-axis forces (Bessler-Etten et al., 2022).
- The lateral mounting of the motor and gearbox adds unilateral bulk and mass. This asymmetry could subtly influence frontal plane balance and kinematics (Jeong et al., 2020).
- Interface compliance and slippage at the pelvic band and thigh cuff, secured by Velcro, could lead to the brace shifting during forceful assistance, pushing the limb segments into abduction or rotation (Lin et al., 2025).
- Uni-axial Actuation Limitations: The current design provides uni-axial actuation (hip flexion/extension). The human hip is a tri-axial joint. Actively assisting one DoF without considering or constraining others can lead to "spill-over" effects. The observed non-sagittal movements might be an unmanaged consequence of this focused assistance (Chen et al., 2020).

These unexpected kinematic outcomes are arguably the most significant area for design implication. While some EMG reduction is positive, inducing undesirable movements in other planes is a concern. Future iterations must address this, potentially through:

- Improved Brace Structure: A custom-designed, more conforming, and rigid base structure with better anatomical joint alignment.
- Multi-DoF Consideration: Incorporating passive or even active constraints/guidance for hip abduction/adduction and rotation.
- Refined Actuation: Moving away from a simple "pulse" to a smoother, more biomimetic torque profile that the user can better integrate, potentially reducing the need for gross compensations.

- Enhanced Human-Robot Interface: Better strapping, custom-molded cuffs, or alternative attachment methods to minimize slippage and ensure forces are applied as intended.

5.4.5 Influence of Passive Brace Mechanics. The kinematic changes observed during the descent phase highlight the impact of the brace as a passive mechanical structure. The reductions in sagittal plane ROM/MAV/RMS at the hip and knee, and the emergence of non-sagittal movements even during descent, suggest that simply wearing the device, with its inherent stiffness, fit, and added components, alters the natural squat pattern (Gregorczyk et al., 2010). This underscores the importance of distinguishing between active assistance effects and passive mechanical effects in the overall results. The increase in TFL VAR during both descent and the full cycle might reflect the muscle's continuous effort to manage the hip's interaction with this passive system.

Chapter 6

General Conclusion, Limitations & Future Work

6.1 General Conclusion

This study presented the conceptualization, development, and evaluation of a hip exoskeleton aimed at assisting users during the ascent phase of squatting. The exoskeleton system combined a commercial hip brace with a custom 3D-printed actuator, strategically triggered at 70° of hip flexion to offer targeted assistance during hip extension. This approach sought to reduce muscular demand on key hip extensors, particularly during the most biomechanically demanding portion of the squat.

Electromyography (EMG) and kinematic analyses confirmed that the device successfully reduced muscle activity in both the Gluteus Medius and Rectus Femoris. This reduction in activity is indicative of effective offloading, aligning with the study's core hypothesis that sagittal-plane-focused support can alleviate muscular strain without introducing external weight or major disruption to the user's movement pattern. Moreover, minor modifications in joint kinematics suggested a redistribution of biomechanical load during ascent, further reinforcing the functional influence of the exoskeleton.

The lightweight and targeted nature of the assistive actuator demonstrates the feasibility of using single-joint, minimalistic designs for biomechanical augmentation. By providing assistance only during a critical portion of the squat, the system maintained a low profile while still achieving measurable benefits. This validates the notion that wearable robotics need not be bulky or complex to be effective—particularly when designed with a deep understanding of human biomechanics.

However, the results also uncovered unintended alterations in movement outside the sagittal plane. Specifically, compensatory movements involving hip abduction and transverse plane instability suggest that isolated sagittal assistance may disrupt overall lower-limb coordination if not properly managed. These findings emphasize the need for holistic control strategies and mechanical designs that consider three-dimensional motion patterns in human movement.

The study confirmed that focused actuation at the hip joint can reduce muscular workload and assist in lower-limb movement during bodyweight squats. The exoskeleton design serves as a promising foundation for the next generation of occupational or rehabilitative assistive devices. Yet, to evolve into a viable real-world solution, further refinement in design, control, and adaptability is essential.

6.2 Limitations & Future Work

Despite the promising findings, several limitations restrict the broader applicability of the study results. The participant pool was limited to a small number of healthy young adult males, which introduces sampling bias and restricts generalizability. Older adults, individuals with musculoskeletal conditions, or occupational users were not included, making it difficult to extrapolate the findings to populations who might benefit most from such assistive technologies.

Furthermore, the study relied on a single-session evaluation conducted in a controlled laboratory environment. Short-term measurements may not reflect long-term adaptations such as motor learning, muscle conditioning, or behavioral habituation. Real-world use cases—characterized by environmental variability, prolonged activity, and repeated daily use—require longitudinal testing to determine the durability and functional relevance of the exoskeleton over time.

On the technical side, the actuator operated exclusively in the sagittal plane and employed an open-loop control strategy. This approach does not account for individual variability in movement patterns or real-time changes in effort or intention. As a result, the assistance may have been mistimed or overly rigid in some instances, potentially contributing to the observed compensatory motions outside the intended plane of support.

Future work should therefore prioritize the integration of closed-loop, adaptive control algorithms. For example, EMG-based intention detection or inertial measurement units (IMUs) could be employed to dynamically adjust torque outputs based on user behavior. This would enable the system to offer assistance that is not only biomechanically synchronized but also personalized to individual needs and environmental demands.

In addition, mechanical redesign should focus on anatomical conformity and stability. Issues such as brace slippage and misalignment can reduce force transmission efficiency and user comfort. Future iterations could explore soft robotics or hybrid designs that blend rigid actuation with flexible, body-conforming materials. Finally, broadening the participant demographics and conducting field-based or longitudinal trials will be critical in validating the exoskeleton's practicality and safety outside of laboratory settings. Through these improvements, the ultimate goal is to create an intelligent, ergonomic, and adaptive hip exoskeleton that supports natural movement and promotes long-term musculoskeletal health



REFERENCES

- Ahsan, M. R., Ibrahimy, M. I., & Khalifa, O. O. (2011, May). Electromyography (EMG) signal based hand gesture recognition using artificial neural network (ANN). In 2011 4th international conference on mechatronics (ICOM) (pp. 1-6). IEEE.
- Barrutia, W. S., Yumiceva, A., Thompson, M., & Ferris, D. P. (2024). Soft tissue can absorb surprising amounts of energy during knee exoskeleton use. *Journal of the Royal Society Interface*, 21(221).
- Bessler-Etten, J., Schaake, L., Prange-Lasonder, G. B., & Buurke, J. H. (2022). Assessing effects of exoskeleton misalignment on knee joint load during swing using an instrumented leg simulator. *Journal of NeuroEngineering and Rehabilitation*, 19(1), 13.
- Blackburn, J. T., & Padua, D. A. (2008). Influence of trunk flexion on hip and knee joint kinematics during a controlled drop landing. *Clinical Biomechanics*, 23(3), 313–319.
- Brice, S. M., Doma, K., & Spratford, W. (2022). Effect of footwear on the biomechanics of loaded back squats to volitional exhaustion in skilled lifters. *The Journal of Strength & Conditioning Research*, 36(10), 2676-2684.
- Chen, B., Zi, B., Qin, L., & Pan, Q. (2020). State-of-the-art research in robotic hip exoskeletons: A general review. *Journal of Orthopaedic Translation*, 20, 4–13.

- Chiaradia, D., Xiloyannis, M., Solazzi, M., Masia, L., & Frisoli, A. (2019). Rigid versus soft exoskeletons: Interaction strategies for upper limb assistive technology. Presented at IEEE ICORR Workshop, RehabWeek.
- De Looze, M. P., Bosch, T., Krause, F., Stadler, K. S., & O'Sullivan, L. W. (2016). Exoskeletons for industrial application and their potential effects on physical work load. *Ergonomics*, 59(5), 671–681.
<https://doi.org/10.1080/00140139.2015.1081988>
- Escamilla, R. F. (2001). Knee biomechanics of the dynamic squat exercise. *Medicine & Science in Sports & Exercise*, 33(1), 127–141.
- Frigotto, M. F., Cardoso, C. A., Santos, R. R. D., & Rodrigues, R. (2019). Gluteus Medius and Tensor Fascia Latae muscle activation levels during multi-joint strengthening exercises. *Motriz: Revista de Educação Física*, 25(3), e101983.
- Gage, K. (2023, November 1). Trigno® - Delsys Europe
- Gallagher, S., & Marras, W. S. (2012). Tolerance of the lumbar spine to shear: A review and recommended exposure limits. *Clinical Biomechanics*, 27(10), 973–978.
- Golabchi, A., Chao, A., & Tavakoli, M. (2022). A systematic review of industrial exoskeletons for injury prevention: efficacy evaluation metrics, target tasks, and supported body postures. *Sensors*, 22(7), 2714.
- Gregorczyk, K. N., Hasselquist, L., Schiffman, J. M., Bense, C. K., Obusek, J. P., & Gutekunst, D. J. (2010). Effects of a lower-body exoskeleton device on metabolic cost and gait biomechanics during load carriage. *Ergonomics*, 53(10), 1263–1275.

- Hamed, M., Salleh, S. H., Noor, A. M., Swee, T. T., & Afizam, I. K. (2012). Comparison of different time-domain feature extraction methods on facial gestures' EMGs. *Progress In Electromagnetics Research*, 1897.
- Hoogenboom, B. J., May, C. J., Alderink, G. J., Thompson, B. S., & Gilmore, L. A. (2023). Three-dimensional kinematics and kinetics of the overhead deep squat in healthy adults: a descriptive study. *Applied Sciences*, 13(12), 7285.
- Jayaraman, C., Mummidisetty, C. K., Jayaraman, A., Pflieger, K., Jacobson, M., Ceruolo, M., . . . Chen, D. (2024). Validity and reliability study of a novel surface electromyography sensor using a well-consolidated electromyography system in individuals with cervical spinal cord injury. *Spinal Cord*, 62(6), 320–327.
- Jeong, H., Haghighat, P., Kantharaju, P., Jacobson, M., Jeong, H., & Kim, M. (2023). Muscle coordination and recruitment during squat assistance using a robotic ankle–foot exoskeleton. *Scientific Reports*, 13, 1363.
- Jeong, H., Haghighat, P., Kantharaju, P., Jacobson, M., Jeong, H., & Kim, M. (2023). Muscle coordination and recruitment during squat assistance using a robotic ankle–foot exoskeleton. *Scientific reports*, 13(1), 1363.
- Jeong, M., Woo, H., & Kong, K. (2020). A study on weight support and balance control method for assisting squat movement with a wearable robot (Angel-suit). *International Journal of Control, Automation and Systems*, 18(1), 114–123.
- Junius, K., et al. (2017). Bilateral, misalignment-compensating, full-DOF hip exoskeleton: Design and kinematic validation. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 25(8), 1501–1512.

- Konrad, P. (2005). The Abc Of Emg. A Practical Introduction To Kinesiological Electromyography, 1(2005), 30-5.
- Kuber, P. M., Alemi, M. M., & Rashedi, E. (2023). A systematic review on Lower-Limb Industrial exoskeletons: evaluation methods, evidence, and future directions. *Annals of Biomedical Engineering*, 51(8), 1665-1682.
- Li, L., Chen, Z., Hong, R., Qu, Y., Gao, X., & Wang, X. (2025). Research Status and Development Trend of Lower-Limb Squat-Assistant Wearable Devices. *Biomimetics*, 10(5), 258.
- Lin, W., Dong, H., Gao, Y., Long, Y., & Dong, W. (2025). A systematic review of locomotion assistance exoskeletons: Prototype development and technical challenges. *Technologies*, 13(2), 69.
- Marzullo, R. A. (2022, August 11). 5 tips on how to draw leg anatomy – tutorial. Ram Studios Comics.
- Matsunaga, N., Okubo, Y., Isagawa, S., Niitsuma, J., Otsudo, T., Sawada, Y., & Akasaka, K. (2021). Muscle fatigue in the gluteus maximus changes muscle synergies during single-leg landing. *Journal of Bodywork and Movement Therapies*, 27, 493-499.
- Nair, D., Rajapriya, R., & Rajeswari, K. (2021). Real-Time Electromyographic Hand Gesture Signal Classification Using Machine Learning Algorithm Based on Bispectrum Feature. In *Lecture notes in electrical engineering* (pp. 559–564).
- Nesler, C., Thomas, G., Divekar, N., Rouse, E. J., & Gregg, R. D. (2022). Enhancing voluntary motion with modular, backdrivable, powered hip and knee orthoses. *IEEE robotics and automation letters*, 7(3), 6155-6162.

- Ottinger, C. (2020, May 15). Hamstrings in the squat. The Muscle PhD.
- Parijat, P., & Lockhart, T. E. (2008). Effects of lower extremity muscle fatigue on the outcomes of slip-induced falls. *Ergonomics*, 51(12), 1873–1884.
- Physiopedia contributors. (n.d.). Gluteus maximus. Physiopedia.
- Ramadurai, R., Kantharaju, P., Jeong, H., & Kim, M. (2023). Predicting the metabolic cost of exoskeleton-assisted squatting using foot pressure features and machine learning. *Frontiers in Robotics and AI*, 10, 1166248.
- Refai, M. I. M., Moya-Esteban, A., Otero, A., & Koopman, A. S. (2024). Benchmarking commercially available soft and rigid passive back exoskeletons for an industrial workplace. *Wearable Technologies*, 5, e6.
- Singh, A., Kashiri, N., & Tsagarakis, N. (2020, November). Design of a quasi-direct-drive actuator for dynamic motions. In *Proceedings* (Vol. 64, No. 1, p. 11). MDPI.
- Straub, R. K., & Powers, C. M. (2024). A biomechanical review of the squat exercise: implications for clinical practice. *International Journal of Sports Physical Therapy*, 19(4), 490.
- Sun, M., Ouyang, X., Mattila, J., Yang, H., & Hou, G. (2021). One novel hydraulic actuating system for the lower-body exoskeleton. *Chinese Journal of Mechanical Engineering*, 34, 31.
- Tu, Y., Zhu, A., Song, J., Zhang, X., & Cao, G. (2022). Design and experimental evaluation of a lower-limb exoskeleton for assisting workers with motorized

- tuning of squat heights. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 30, 184–193.
- Urrea, C., & Agramonte, R. (2023). Improving Exoskeleton Functionality: Design and Comparative Evaluation of Control Techniques for Pneumatic Artificial Muscle Actuators in Lower Limb Rehabilitation and Work Tasks. *Processes*, 11(12), 3278.
- Van den Tillaar, R. (2019). Effect of descent velocity upon muscle activation and performance in two-legged free weight back squats. *Sports*, 7(1), 15.
- Wang, Z., Wu, X., Zhang, Y., Chen, C., Liu, S., ... Ma, Y. (2021). A semi-active exoskeleton based on EMGs reduces muscle fatigue when squatting. *Frontiers in Neurorobotics*, 15, 625479.
- Wattananon, P., Kongoun, S., Klahan, K., Silfies, S. P., Gilliam, J. R., & Richards, J. (2025). Trunk kinematics and motor unit behavior during different loads and speeds in individuals with and without aberrant movement patterns during active forward bending: A cross-sectional study. *PloS one*, 20(4), e0321084.
- Wei, W., Zha, S., Xia, Y., Gu, J., & Lin, X. (2020). A hip active assisted exoskeleton that assists the semi-squat lifting. *Applied Sciences*, 10(7), 2424.
- Wu, G., Siegler, S., Allard, P., Kirtley, C., Leardini, A., Rosenbaum, D., . . . Stokes, I. (2002). ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion—part I: ankle, hip, and spine. *Journal of Biomechanics*, 35(4), 543–548.

Xu, L., Liu, X., Chen, Y., Yu, L., Yan, Z., Yang, C., Zhou, C., & Yang, W. (2023).

Reducing the muscle activity of walking using a portable hip exoskeleton based on human-in-the-loop optimization. *Frontiers in Bioengineering and Biotechnology*, 11, 1006326.

Zawadka, M., Smolka, J., Skublewska-Paszkowska, M., Lukasik, E., & Gawda, P. (2020).

How are squat timing and kinematics in the sagittal plane related to squat depth?. *Journal of sports science & medicine*, 19(3), 500.

Zhong, B., Cao, J., McDaid, A., Xie, S. Q., & Zhang, M. (2020). Synchronous position and compliance regulation on a bi-joint gait exoskeleton driven by pneumatic muscles. *IEEE Transactions on Automation Science and Engineering*, 17(4), 2162–2166.

Zhou, S., Yin, K., Fei, F., & Zhang, K. (2019). Surface electromyography–based hand movement recognition using the Gaussian mixture model, multilayer perceptron, and AdaBoost method. *International Journal of Distributed Sensor Networks*, 15(4), 1550147719846060.

Zhu, J., Jiao, C., Dominguez, I., Yu, S., & Su, H. (2022). Design and backdrivability modeling of a portable high torque robotic knee prosthesis with intrinsic compliance for agile activities. *IEEE/ASME Transactions on Mechatronics*, 27(4), 1837-1845.