INFLUENCE OF DUAL COGNITIVE-MOTOR DEMANDS ON KNEE MECHANICS RELATED TO ACL INJURY IN RUN-AND-CUT TASKS

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ABSTRACT

Among females, non-contact anterior cruciate ligament (ACL) injuries remain a critical concern in dynamic sports involving rapid decelerations and accelerations, and directional changes. Although injury mechanisms and risk factors are well-documented, the role of cognitive demands during movement execution remain underexplored. Real-world sport performance requires athletes to manage environmental cues, maintain and manipulate information in working memory, and make rapid decisions while simultaneously executing movement. Concurrent cognitive and motor demands are likely to compete for attentional resources, which may influence movement coordination during unanticipated conditions relevant to ACL injury-risk mechanics. Therefore, the purpose of this dissertation was to examine the effects of cognitive load and movement anticipation on knee joint biomechanics during a 45° sidestep run-and-cut task.

Chapter 4 examined the influence of cognitive load and anticipation on knee joint biomechanics and EMG activity using linear mixed-effects models. A significant interaction between anticipation and cognitive load was observed for anteroposterior ground reaction forces. Anticipation also increased peak knee extensor moments, whereas unanticipated cuts involved reduced braking forces and greater knee flexion at initial contact and peak, indicating increased joint excursion. Chapter 5 quantified cognitive-motor dual-task costs to characterize patterns of cognitive-motor interference. No significant differences in cognitive performance were observed across conditions. However, in the anticipated condition, higher dual-task effect (DTC) scores were positively associated with peak knee external rotation and adduction moments, as well as anteroposterior ground reaction forces, whereas in the unanticipated condition, DTC was associated with both knee joint moments and flexion angle. Participants demonstrated

heterogeneity in attention trade-off strategies, with some prioritizing motor stability at the expense of cognitive accuracy.

Overall, the findings reveal that cognitive load and anticipation shape movement through distinct but interacting attentional mechanisms, linking cognitive–motor processes to variability in knee joint control relevant to ACL injury risk. This integrative approach advances understanding of how attention and biomechanics interact during complex movements. Future work should apply these methods to more demanding cognitive tasks and clinical populations to enhance ecological and practical applications.

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NOMENCLATURE

ACL	Anterior cruciate ligament
ACL-R	Anterior cruciate ligament reconstruction
BW	Body weight
EMG	Electromyography
MVIC	Maximum voluntary isokinetic contraction
deg/sec	Degrees per second
F	Force
GRF	Ground reaction force
N	Newtons
kg	Kilogram
m	Meters
Nm/kg	Newton-meters per kilogram
COG	Cognitive load condition
NOCOG	No cognitive load condition
AN	Anticipated condition
UN	Unanticipated condition
NOCOG_AN	No cognitive load, anticipated condition

NOCOG_UN	No cognitive load, unanticipated condition
COG_AN	Cognitive load, anticipated condition
COG_UN	Cognitive load, unanticipated condition
LMM	Linear mixed model
DTC	Dual task cost
CMI	Cognitive motor interference
Q/H ratio	Quadriceps-to-hamstring ratio

Chapter 1: Development of the Problem

Background and Rationale

The anterior cruciate ligament (ACL) is essential for maintaining lower extremity stability at the knee joint. Injuries to this ligament often occur in dynamic sports that require concurrent cognitive processing (e.g., decision-making) and precise motor control [1-3]. Approximately 55%-80% of ACL injuries are classified as non-contact, typically resulting from actions of rapid deceleration or sudden changes in direction [4,5]. Extensive research has progressed the understanding of both risk factors and etiology behind non-contact ACL injuries [6-8]. Broadly, risk factors (i.e., characteristics) are categorized into intrinsic and extrinsic factors. Notably, females exhibit higher incidence rates compared to males, attributable to differences in anatomical structures, neuromuscular control, and hormonal influences [8-10]. Biomechanical factors such as improper joint alignment, neuromuscular imbalances, or external forces are attributed to underlying mechanisms of injury. While extensive research on these risk factors and mechanisms has been conducted, the influence of cognitive demands on injury mechanics during dual task scenarios that replicate sport conditions remains under explored.

Emerging evidence suggests that neurocognitive factors, such as reaction time, processing speed, and memory, may influence ACL injury risk [11]. A case control study found that athletes who later sustained non-contact ACL injury exhibited lower baseline neurocognitive test scores compared to controls [12]. In line with this, neurocognitive deficits have been associated with injury risk patterns in healthy individuals [13-16]. Furthermore, risk of injury is particularly heightened in agility sports that involve rapid deceleration and directional changes, with cutting identified as a primary action linked to non-contact injuries [5,17-19]. Notably, these movements often occur in dual task scenarios, where athletes must execute a motor task (e.g., cutting) while managing cognitive demands [20]. In sports, cognitive tasks include tracking opponents, processing on-the-field actions, play recalls, and split-second decision making, all of which are further influenced by the available time to react. The added cognitive load can interfere with the athlete's ability to maintain optimal motor control, further contributing to risk of injury.

In the context of ACL injury risk, several studies have implemented dual task paradigms involving landing and cutting – movements commonly linked to ACL injuries due to rapid decelerations and abrupt directional changes [21,22]. During dual task landing conditions, researchers have observed biomechanical patterns associated with increased ACL injury risk, such as decreased knee flexion angles, greater peak ground reaction forces [2,23-25], and increased tibial rotation angles [26]. Similar trends have been reported during unanticipated cutting tasks [15,27,28]. During cutting, athletes must monitor and react to sudden changes in both their own and their opponent's movement and make rapid cognitive decisions. However, comparing findings across studies is rather difficult due to the inconsistencies in the dual task paradigms implemented, particularly in how both cognitive and physical tasks are integrated.

True cognitive-motor dual tasks necessitate the simultaneous execution of both mental and physical activities [29]. Unanticipated cutting involves both physical movement and quick decision-making responses, broadly classifying it as a dual task [30]. Despite extensive research on lower extremity biomechanics during cutting, the measurement and definition of cognitive load remain inconsistent [12,15,16]. In many studies, cognitive load is inferred based on task complexity or performance outcomes, however, the mental effort involved is not directly quantified. Without standardized measures of these data, it becomes difficult to assess its true impact on motor performance and injury risk.

Recognizing the variability in existing paradigms, Wollesen, Wanstrath, van Schooten and Delbaere [31] proposed a taxonomy for cognitive-dual tasks, categorizing them based on the cognitive processes involved. Specifically, this taxonomy distinguishes tasks involving simple-reaction time demands, decision-making (controlled processing), visuospatial processing, mental tracking (e.g., counting backwards or verbal fluency), working memory, and discrimination tasks (Go/No-Go). This taxonomy provides a structured framework for systematically identifying and addressing the cognitive demands in dual task scenarios, offering a more standardized approach for selecting and comparing cognitive-motor dual-task conditions by making explicit what type of cognitive load is imposed. Without inclusion of well-defined cognitive components, studies may overlook the full impact of cognitive load on movement patterns and performance outcomes. In addition, Plummer, Eskes, Wallace, Giuffrida, Fraas, Campbell, Clifton and Skidmore [32] have developed standard measures of cognitive load and interference, offering valuable tools for assessing the cognitive demands in clinical dual task scenarios. However, stemming from gait applications [33,34], these methods have yet to be consistently applied in ACL-related studies. Thus, a gap remains in integrating these measures with motor tasks, such as cutting, to fully understand how cognitive demands influence performance and ACL injury risk.

Statement of the Problem

Athletes often perform complex motor tasks under high cognitive demands, such as making quick decisions while executing precise movements during cutting. However, the current understanding of cognitive demands and complexity on performance and lower extremity mechanics during cutting remains under explored. Research has largely focused on biomechanical factors such as joint angles and moments [14,35,36], but the interaction of cognitive load—such as decision making and working memory demands—on movement patterns has received less attention. Added cognitive factors may alter movement patterns and disrupt neuromuscular control, leading to decreased performance in one or both tasks and elevate injury risk [15,26,28,37,38]. Given the high prevalence of ACL injuries, there is a pressing need to understand the interaction of cognitive and biomechanical factors that contribute to injury risk [20]. Measuring cognitive load alongside biomechanical factors in a dual task paradigm can enhance our understanding of injury risk, ultimately leading to more informed and effective prevention and rehabilitation protocols.

Statement of Purpose

The purpose of this dissertation was to determine how cognitive load influences lower extremity mechanics during cognitive-motor dual tasks. Thus, the following specific aims were proposed to accomplish this goal. Specific aim 1 was to examine the main effect of working memory on lower extremity biomechanics during anticipated and unanticipated cutting maneuvers in healthy females. Specific aim 2 was to examine the main effect of cognitive load which was influenced by placing a demand on working memory. on muscle activity during anticipated and unanticipated cutting maneuvers in healthy females. Specific aim 3 was to examine the interactions between movement anticipation and working memory in healthy females.

The secondary purpose was to examine dual task cost effects within the framework of attentional resource theories. Differences in dual task interference across conditions help to clarify the extent to which cognitive load impacts motor performance. This study aimed to quantify cognitive-motor interference, shedding light on how the brain prioritizes cognitive versus motor tasks.

Research Hypotheses

Previous research suggests that ACL injury risk is elevated during dual task scenarios, particularly during landing maneuvers [2,3,24]. Therefore, it was hypothesized that greater degrees of ACL injury mechanics would be observed under dual cognitive-motor conditions. Specifically, cognitive conditions were expected to result in decreased knee flexion angles, increased knee abduction angles, increased peak vertical GRF, and greater external knee abduction moments. Additionally, individual muscle activation and quadricep-to-hamstring co-activation were expected to be reduced during working memory conditions, compared to run-and-cut trials without an additional cognitive task.

Building on this, the interaction between increased cognitive demand and unanticipated movement was expected to impair motor performance. This interaction was hypothesized to result in the most significant changes in joint mechanics and muscle activation—such as decreased knee flexion angles, increased knee abduction angles, increased peak vertical GRF, and delayed quadriceps-hamstring muscle activation—compared to anticipated cutting. Thus, incorporating a working memory task during unanticipated cutting was expected to result in unfavorable biomechanics which are linked to higher ACL injury risk.

Cognitive performance measurements were attained to assess how cognitive load and complexity impact performance of both tasks [39]. It was hypothesized that higher cognitive demands, indicated by inclusion of working memory tasks, would lead to altered joint mechanics, ultimately elevating injury risk. Furthermore, these data provide insight into the specific types of cognitive interference, allowing for a more comprehensive understanding of their impact on both motor and cognitive performances. By classifying these interferences, we can better identify the factors that affect lower extremity mechanics and contribute to ACL injury.

Independent Variables

- Cognitive load: (Working memory task and no working memory task)
- Anticipation: (Unanticipated and Anticipated)

Dependent Variables

- Initial contact knee angles
    - Knee flexion/extension angle
    - Knee abduction/adduction angle
    - Knee internal/external rotation angle
- Peak knee angles
    - Knee flexion/extension angle
    - Knee abduction/adduction angle
    - Knee internal/external rotation angle
- Peak knee moments
    - Knee flexion/extension moment
    - Knee abduction/adduction moment
    - Knee internal/external rotation moment
- Peak vertical ground reaction force
- Peak mediolateral ground reaction force

- Peak anteroposterior ground reaction force
- Mean Vastus Medialis Electromyography Activation Signal
    - Pre-stance
    - Stance Phase
- Mean Medial Hamstring Electromyography Activation Signal
    - Pre-stance
    - Stance Phase
- Quadriceps-to-hamstring (Q/H) ratio
    - Pre-stance
    - Stance Phase

Significance of the Study

Given the limited research on true cognitive-motor dual tasks in cutting, this study aimed to advance research in this area by implementing and demonstrating a well-defined dual task paradigm. Recognizing the interplay of neurocognitive performance and movement patterns during cognitive-motor dual tasks can help provide a greater understanding of how to address and reduce injury risk and potentially develop more effective rehabilitation protocols.

Limitations of the Study

- Participants in the proposed study had various sports backgrounds and cutting experience.
- The sample consisted of healthy, physically active young females, which may limit generalizability to other populations (e.g., clinical, youth, or elite athletes).
- The relatively small sample size (≈18 participants) was appropriate for within-subject analyses but may limit external validity and detection of smaller effects.

- Participants may have exhibited a learning effect for unanticipated cutting; however, this was expected to have a negligible impact on the study outcomes.
- Laboratory-based cutting tasks may not fully replicate the environmental complexity or movement variability present in real sport contexts.
- The 45° sidestep cut reflects a single movement pattern and may not represent other high-risk maneuvers (e.g., pivoting, landing).
- The cognitive task (digit recall) represents one form of working-memory demand; results may differ with other cognitive domains (e.g., decision-making, inhibition, or visual tracking).
- Anticipation was manipulated through known vs unknown cues, which may not capture the range of perceptual-cognitive processes used in reactive sport situations.
- EMG normalization to maximum voluntary isometric contractions (MVIC) introduces variability related to participant effort and electrode placement.
- EMG analysis included only the vastus medialis and medial hamstring, limiting the neuromuscular evaluation of other knee stabilizers (e.g., biceps femoris, gastrocnemius, and vastus lateralis).
- Cognitive performance was assessed via accuracy and dual-task efficiency metrics, which may not capture all aspects of attentional allocation or mental workload.
- Findings reflect acute responses to cognitive load rather than long-term adaptations or training effects.
- The cognitive task used to induce load was relatively simple and may not have imposed a sufficiently high or ecologically valid level of cognitive demand.

Delimitations of the Study

- Participants' ages were between 18 to 30 years old.
- Participation was restricted to females only.
- Any participant who experienced a lower extremity injury within the past six months was excluded.
- Any participant experiencing pain on or during the day of the testing was excluded.
- The motor task selected was a 45° sidestep cut, chosen for its relevance to ACL injury mechanisms and repeatability under laboratory conditions.
- Baseline cognitive testing was included to provide an individualized comparison for each participant's dual-task cognitive performance.

Assumptions of the Study

- Participants provided accurate information on their health history forms concerning lower extremity injury history and weekly activity levels.
- The twelve-camera infrared motion capture system (Vicon Motion Analysis, Inc., Centennial, CO, USA) and force platforms (Advanced Mechanical Technology, Inc., Watertown, MA, USA) were accurately calibrated before each data collection.
- Placement of electromyography electrodes and retroreflective markers were consistent across participants.
- The EMG processing methods used were sufficient to maintain accurate and reliable representations of muscle activity and movement patterns.
- Statistical assumptions of the linear mixed-effects models (normality, homogeneity of variance, independence) were sufficiently met for valid inference.

Operational Definition of Terms

ACL loading: increased tensile forces and strain upon the ACL, increasing elongation of the ligament.

Cognitive-motor dual task: simultaneous engagement of one activity that requires mental processing and another that involves physical movement, each with distinct and separate goals.

Cognition: mental processes involved in organizing and understanding external sensory input. Central integration of these processes is attention – the ability to allocate cognitive resources to the tasks presented.

Cognitive load: the degree of attentional allocation required to manage the task presented in both single- and dual-task conditions. In the present study, cognitive load was induced by a concurrent working memory task in which participants encoded and recalled digit sequences.

Working Memory: the temporary storage and manipulation of information for immediate use, measured through performance on a digit recall task completed under single-and dual-task conditions.

Joint moments: the way a force causes rotation about an axis. In this study, internal joint moments are defined as the net rotational effect produced by the combined agonist and antagonist muscle forces about a joint.

Joint kinematics and kinetics will be defined using the right-handed Cartesian coordinate system (x-y-z).

Hip: flexion (+), extension (-); adduction (+), abduction (-); internal rotation (+), external rotation (-)

Knee: flexion (-), extension (+); adduction (+), abduction (-); internal rotation (+), external rotation (-)

Ankle: dorsiflexion (+), plantarflexion (-); inversion (+), eversion (-)

Motor task: physical voluntary and purposeful movement.

Noncontact ACL injury: ligament sprain or tear that is sustained in the absence of direct player contact to the knee. These injuries often occur during unanticipated athletic movements such as cutting, single-leg landing, or rapid deceleration.

Recreationally active: according to ACSM guidelines, individuals who engage in moderate-intensity aerobic activity at least five days per week for a minimum of 30 minutes per session, or vigorous intensity aerobic activity three days per week for a minimum of 20 minutes per session.

Chapter 2: Review of the Literature

Introduction

Injuries to the ACL are highly costly, both financially and in terms of the impact on an athlete's career. These injuries often require extensive rehabilitation, and, in many cases, surgical intervention. Despite these efforts, some athletes may struggle to return to their previous level of performance, and in certain cases, the injury can be career-ending [40,41]. The psychological toll of such injury can also be considerable, affecting an athlete's mental health and overall well-being [42,43]. Extensive research has established a thorough understanding of the primary injury mechanism and attributing risk factors [9,44,45]. However, the role of cognitive demand pertaining to dual tasks remains under-explored.

For high-demand sports that require precise movement coordination and the ability to rapidly process cognitive loads, athletes must simultaneously manage stimuli from both tasks. Athletes under high cognitive demands may exhibit compromised neuromuscular control, elevating their risk of injury [13]. Sensory receptors of the ACL are key contributors for information processing of joint position and reflexes [46,47]. However, increased demands in both cognitive and motor tasks influence the likelihood of both physical and cognitive overload. Thus, ACL injury risk may be elevated due to altered performance.

The purpose of this dissertation is to determine how cognitive-motor dual tasks contribute to ACL injury mechanics. This chapter reviews literature on the ACL and dual tasks, starting with an exploration of ACL anatomy, factors contributing to ACL loading, and injury mechanisms. It then provides an overview of dual task theories, frameworks, and paradigms, leading into their application to biomechanics. Recognizing the implications of dual task performance is crucial for understanding its impact on clinical populations. Given the prevalence of ACL injuries, there is a pressing need to understand the cognitive and biomechanical factors and interactions that contribute to these injuries.

ACL Overview

ACL Anatomy

The ACL is a critical structure of the knee joint, originating from the posteromedial lateral femoral condyle and inserting in the tibial anterior intercondylar area [48]. It is primarily classified into two bundles – the anteromedial and posterolateral bundles – each with distinct roles in knee stability [49,50]. These bundles are designed to respond to mechanical loads, with the anteromedial bundle primarily controlling anterior tibial translation and the posterolateral resisting rotational forces [51-53]. However, given the physiological continuum that enables both bundles to work synergistically, we will consider the ACL as a single functional unit [53,54].

Ligament length varies from 22 to 41 mm, with an average of 33 mm, and its width spans from 7 to 12 mm [51,55-57]. Optimal length and width are essential for maintaining joint stability and withstanding stress during high-impact movements that could lead to injury [58-60]. While length influences range of motion by detecting the amount of tension or stretch, width enables the ligament to resist tensile forces. A greater width allows the ligament to distribute applied stress across its cross-sectional area, enhancing the capacity to withstand high loads [61]. Sex differences in ligament structure (i.e. cross-sectional area) have shown greater risk of ACL injury in females [57,61]. Morphological variations stem from personal characteristics and genetics (e.g., hormones and anthropometrics), as well as biomechanical factors such as collagen composition and ligament stiffness, tensile strength, and elasticity. Overall, these structural properties contribute to function, while mechanical properties are determinants of its performance.

The ACL is predominantly composed of type I collagen fibers (~90%), which provides high tensile strength, allowing it to resist forces encountered during dynamic activities such as jumping, cutting, and landing [48]. The remaining composition (~10%) includes type III collagen fibers, elastin, fibroblasts, and fibrocartilage, each contributing to the ligament's elasticity, flexibility, and ability to absorb and distribute stress[62]. Though less abundant, type III collagen fibers are essential for the ligament's ability to adapt to mechanical demands. Elastin further enhances flexibility, enabling it to return to its original shape after deformation. Fibroblasts aid in maintaining tissue structure and repair, while fibrocartilage provides additional resilience. Moreover, the interlaced network of collagen fibrils reinforces the ligament's structural integrity under multiaxial stress or load [63,64]. This intricate composition of fibers and cells enables the ACL to function effectively in both stabilizing the knee and facilitating movement, a role further enhanced by encapsulated mechanoreceptors.

ACL Function

Often, movement is a result of unconscious activation governed by the neuromuscular system [65]. Knee joint sensory awareness is governed by the peripheral nerves that innervate the ACL and transmit proprioceptive information to the central nervous system[47,64,66]. These nerves contain mechanoreceptors that detect changes in joint position, providing the brain with real-time feedback about the knee joint position and range[67,68]. These mechanoreceptors, embedded within the ligament's collagen fibers govern proception, allowing for movement control [69]. At the time of injury, mechanoreceptors are abruptly disrupted, diminishing the knees' appropriate reflexive response.

The provided sensory feedback is crucial for regulating knee movement and stability. Under normal conditions, dynamic regulation allows the ACL bundles to work in tandem and adjust their activity according to joint position. At lower knee flexion angles (0°-30°) stability is primarily provided by the posterolateral bundle, while the anteromedial bundle becomes more involved in controlling knee flexion at larger angles (>60°) [51,53,55,70,71]. Both bundles become taut at full knee extension, making this a high injury risk position, especially when combined with large joint reaction forces, experienced during dynamic activities [72,73].

While the ACL's function in joint stabilization and movement coordination has been discussed through its structural function and proprioceptive feedback, its effectiveness in these tasks is closely tied to its mechanical properties. Understanding the ligament's response to multiplanar loading is essential for understanding injury risk.

ACL Loading

Loading of the ACL refers to the tension or strain placed on the ligament due to external forces. The ligament's response to stress–internal force per unit of area–results in strain, or tissue deformation. Several factors influence ligament loading, such as knee position, joint kinematics, muscle activations, and the direction and magnitude of applied external forces. The main types of loads at the knee joint that generate high ACL forces include anterior translation of the tibia relative to the femur, rotational loads, compressive, and shear forces [74,75]. Each of these loads plays a distinct role in ACL loading and can contribute to injury depending on the specific movement and joint position [76]. To better understand ligament loading, it is essential to examine tissue properties, such as stiffness, tensile strength, and elasticity and their role in ligament response to forces.

Extensive research has investigated the ACL's structural integrity under various loading conditions, via in-vivo, in-vitro, and simulation studies [77-79]. Woo, Hollis, Adams, Lyon and Takai [80] conducted an in vitro study examining the influence of age on ligament mechanical properties, noting a decline with increasing age. Loading in anatomical orientation, the 22-35 year age group exhibited a mean ultimate tensile load of 2160 ± 157 N, which was 44% and 328% higher than the 40-50 year and 60-97 year age groups, respectively [80]. Among the younger age group, ultimate loads were 35% lower under tibial oriented loading (i.e. in line with the femoral and tibial ligament insertion sites), ultimate loads were 35% lower than those for the anatomical orientation [80]. Ligament stiffness, measured as the slope of the load elongation curve, was 242 ± 28 N/mm in anatomical orientation and 218 ± 27 N/mm in tibial orientation for the younger age group [80]. Corroborating and expanding on these findings, Chandrashekar, Mansouri, Slauterbeck and Hashemi [77] further explored structural and mechanical properties by sex, revealing that males exhibited higher thresholds at failure than females. During in vitro cadaveric testing (mean age 36.8 years), males reached failure at 1818 ± 699 N, compared to 1266 ± 527 N for females, who exhibited lower strain (8.3%) and stress (14.3%), and a 22.5% decrease in modulus of elasticity at failure [77]. Coinciding with these findings, females had lower ligament length, area, and volume, suggesting these structural differences may contribute to the observed mechanical property variations and higher injury susceptibility to injury in female ACLs compared to males. Notably, structural properties decline after maturation due to age-related changes in collagen fiber structure and overall tissue elasticity, elevating the risk of injury

(Ahmad et al., 2006; Hollis et al., 1991). Additionally, mechanical, and neuromuscular properties work in tandem to manage the ligament and joint loading.

Assessing movement patterns helps provide insight into how the knee responds to forces during dynamic activities. Factors that influence ACL loading include poor neuromuscular control, characterized by increased knee abduction moment, along with increased joint laxity are often surrogate measures of ACL loading [81,82]. These factors contribute to excessive strain and load on the ACL, particularly during dynamic movements involving deceleration of the body combined with sudden directional changes, increasing the risk of injury [83]. Considerable research has demonstrated the primary function of the ACL is to resist excessive anterior translation of the proximal tibia relative to the distal femur, with a secondary role in restraining excessive knee valgus and internal tibia rotation [49,60,74,84]. In response to quasi static loads, anterior tibial shear force significantly increased ACL strain and force [49,59,85]. In vitro simulations of anterior tibial loads showed that at 30° of knee flexion, ligament force was approximately equal to the applied force (100 N) and increased to 150% at full extension [85]. Furthermore, in ACL-intact cadaveric knees, ligament transection resulted in increased anterior tibial translation without changing in applied force [49,60,86]. Expanding on these findings, the Ball, Stephen, El-Daou, Williams and Amis [86] reported the ACL contributed 63-78% of the total restraint from 0°-90° of knee flexion. At lower knee flexion angles, findings show that quadriceps forces induce anterior tibial translation [87-90]. Hirokawa, Solomonow, Yun, Lou and D'Ambrosia [88] observed isolated quadriceps loading increased tibial displacement at knee flexion angles between 0° and 80°, with peak displacement of 8 mm occurring at 30° for a 12 kg load. Within this range, passive internal tibial rotation occurred and peaked at 15° of knee flexion [88]. The observed kinematic coupling demonstrates how quadriceps loading at low knee flexion angles may place greater strain on the ACL, which must restrain both anterior displacement and rotational forces.

During rapid decelerations and accelerations, and sudden directional changes, the ACL experiences increased loading demonstrating the impact of multiplanar forces. This is further supported by, Kittl, El-Daou, Athwal, Gupte, Weiler, Williams and Amis [91], who, during simulated pivot shift tests, found that the ACL's contribution to restraining internal rotation torque varied with knee flexion. At full extension, the ACL contributed 19.7 ± 7.9%, while at 60° and 90° of flexion its contribution was negligible, around 1% [91]. In line with this, ACL strain increased with combinations of internal tibial torque and knee valgus moments (7.0%±4.1%), when compared to external tibial torque (2.4% ± 3.2%) [92]. Furthermore, the combination of anterior tibial force and internal tibial torque resulted in greater ACL loading when in knee extension [85]. Applying a varus moment during knee extension increased ligament force, while a valgus moment in a flexed position had the same effect [85]. These studies provide further support that ligament loading is influenced by multiplanar kinematics, particularly when internal rotation and valgus moments are combined. These forces place greater strain on the ACL near knee extension due to its role in stabilizing the joint. By facilitating knee flexion and controlling tibial rotation, the biceps femoris helps stabilize the knee during high-stress activities, reducing excessive knee valgus and internal rotation. This muscle's role is particularly crucial during dynamic movements such as sidestep cutting [93] and single-leg landing [94], acting as an agonist for knee flexion and external rotation of the tibia. These contributions highlight the influence of muscle forces on ACL loading.

Muscles play a crucial role in either contributing or opposing knee joint load and is often dependent on the movement and knee joint angle [76]. As discussed, ACL strain is highest when the knee is near full extension (30°-0°), with some studies reporting a range of 60°-0°[4,85,95]. Within this range, the ACL limits anterior tibial translation, while the quadriceps function as extensors, pulling the tibia forward, increasing anterior shear force and ligament demand. However, as the knee moves into deeper flexion (60°-120°), ligament tension is reduced [59]. At high knee flexion angles, the quadriceps' contribution to ACL loading decreases due to the changing angle between the patella tendon and tibia, further reducing the muscle force on the tibia [76,87,88]. Meanwhile, the hamstrings have a greater mechanical advantage in producing posterior shear force, further unloading of the ACL [96]. Overall, it is widely recognized that the quadriceps and hamstrings play crucial roles in ligament loading [97,98], particularly through co-contractions [90,99-103]. Additionally, a modeling simulation study by Maniar, Schache, Sritharan and Opar [93] found that the medial gastrocnemius contributed to anterior shear force during the weight acceptance phase of side-step cutting. During this phase, an external rotation knee joint reaction moment was driven by the quadriceps and soleus [93]. Furthermore, research has also identified weak gluteal muscles as contributors to dynamic knee valgus collapse, a high injury risk position [104,105]. Overall, the studies presented thus far provide insights into how the interaction between muscle forces and joint movements leads to complex loading patterns, which can place varying levels of strain on the ACL. Multi-planar loading is particularly evident during dynamic activities such as cutting, jumping, and landing, which demand muscle coordination and joint stabilization [106]. During injury instances, the forces on the ligament exceed its structural capacity, resulting in damage or rupture.

ACL Injury Overview

Epidemiology

Despite its robust structure, ACL injuries are highly prevalent, particularly in high-dynamic sports that involve rapid deceleration, pivoting, and directional changes [54,107]. In the United States, approximately 200,000 injuries occur each year, with an estimated annual incidence of 120,000 [54]. Notably, non-contact injuries– those that occur without direct impact– account for 55-80% of the total injury rate [4,5,108]. These injuries are particularly concerning as they underscore the critical role of biomechanics and neuromuscular control in maintaining knee stability. Moreover, highlighting the injury prevalence, data from the NCAA Injury Surveillance Program between 2014 and 2019 reported 729 ACL injuries, resulting in an injury rate of 0.86 per 10,000 athletes [6]. Ligament tears were more common during competition (58.6%) with an incidence rate of 2.47, compared to 0.45 in practice. Female athletes had a higher overall rate of 0.95 per 10,000 athletes, with soccer recording the highest incidence rate at 2.60. Comparatively, males had an overall rate of 0.80, with football reporting the highest incidence (1.44) [6]. In sex-comparable sports, females recorded an incidence rate approximately here to four times higher in softball and basketball than their male counterparts [6]. While these data focus on college-aged athletes, it’s essential to note that age is a recognized risk factor, with younger athletes often at greater risk of primary injuries [8]. Given the high prevalence of non-contact injuries, understanding factors that increase athletes’ injury risk is essential.

Risk Factors

Injury risk factors are typically classified into two categories: intrinsic and extrinsic. Intrinsic factors refer to individual characteristics such as anatomical structures, neuromuscular function, and hormonal variations[7]. On the other hand, extrinsic factors involve external influences including the type of activity type, environment in which it takes place, and equipment used [7]. Overall, each factor can influence movement patterns, dynamic control, and ultimately impact injury risk.

Intrinsic Risk Factors. Structural knee geometry features, ACL volume, and knee laxity are of relative importance for understanding knee joint stability [109]. Variations in bony morphology include the femoral condyles, intercondylar femoral notch, and proximal tibia [110-113]. Differences in the femoral intercondylar notch width (i.e. housing of the ACL) may place physical constraints on the ligament during knee rotation or extension [44,109,114,115]. ACL injuries have been associated with a narrow intercondylar notch width, typically between 10.4 and 22.5 mm, compared to 15.3-234 mm in ACL-intact knees [110,116]. Furthermore, while ligament movement can be restricted by a narrow intercondylar notch, the tibial plateau angle greatly influences joint stability by impacting compressive and shear contact forces on the medial and lateral aspects of the tibiofemoral joint [117,118].A steeper sagittal tibial slope allows for increased anterior tibia displacement relative to the femur, placing additional stress on the ACL and adding strain to the ligament, particularly during activities involving sudden deceleration or changes in direction [109]. Alongside this, a steep frontal slope has been associated with increased injury risk, particularly with an increased lateral slope (i.e. outer edge tilt), reducing stability of valgus forces at the knee [116,119]. ACL-injured patients were found to have a steeper, posterior-inferior-directed lateral tibial slope compared to uninjured individuals, supporting its role as a structural risk factor for ACL injury [120]. Additionally, shallower tibial plateau depths were present in female patients, relative to uninjured females [120]. Anatomical differences in lateral tibial slope and depth increase susceptibility to ligament strain, particularly during movements that exhibit high anterior and rotational forces to the knee.

Ligament strength is heavily governed by its cross-sectional area, with a larger area indicating greater strength due to a higher collagen fiber concentration. This structural support is particularly crucial during high dynamic activities like jumping, pivoting, or cutting, where the ACL endures high tensile and torsional forces. ACL cross-sectional area varies in length, with its widest point approximately 38% from the tibial insertion for both males and females [61]. However, greater peak cross-sectional area is typically observed in males (49.0 mm^2) compared to females (41.5 mm^2) [61], which may partly explain the higher ACL injury risk observed in females [121]. Additionally, MRI images have shown an inverse relationship between ACL volume and injury risk, with smaller volumes correlating with higher injury susceptibility [122]. Larger ligament cross-sectional area allows for more even distribution of forces, reducing localized stress points that could lead to tissue failure.

Hormone profiles of particular interest in ACL injury risk research include estrogen, progesterone, and relaxin [121,123-125]. Estrogen is studied for its influence on collagen synthesis and ligament tensile strength. Progesterone is believed to counteract some of estrogen's effects, potentially stabilizing ligament properties. Relaxin, often linked to pregnancy, may increase ligament laxity at elevated levels, thereby heightening the risk of ligament injury. Estrogen and relaxin both effect ligament tensile properties, with estrogen receptors present in ACL fibroblast [123,126]. Hormonal fluctuations corresponding with the menstrual cycle are posited to have direct effects on the neuromuscular system and dynamic joint control [9,127]. Previous research is suggestive that oral contraceptives may contribute to reduced injury rates [127-129]. Comparisons between females on contraceptive pills and controls, revealed no significant changes in muscle strength or fatigue throughout the menstrual cycle [125]. Contrarily, females not on the contraceptive pill exhibited increased quadricep and handgrip strength, and more fatigable during ovulation, compared to the other menstrual phases [125]. Furthermore, during arthrometer assessments, female athletes using oral contraceptives demonstrated decreased anterior-posterior tibiofemoral translation, indicating greater passive knee stability [9,124]. Whilst substantial research has explored the effects of oral hormonal contraceptive use on ACL injury incidence and related risk factors, it is crucial to note considerable limitations remain, making it challenging to draw definitive conclusions. Homogeneous hormone levels vary greatly between individuals, adding complexity to the findings, and the lack of longitudinal data further limits our understanding of the long-term impacts on joint health and ACL injury risk.

Additionally, neuromuscular properties influence joint mechanics. Knee joint stability relies heavily on coordinated activation of the quadriceps and hamstrings [85,130,131]. The quadriceps primary role is to extend the leg, whereas the knee flexion is carried out by concentric hamstring contractions. Neuromuscular deficits, such as muscle imbalances and fatigue, can disrupt this coordination, timing, and responsiveness, impairing dynamic knee control [65]. Overly dominant quadricep contractions place greater tensile forces on the knee joint, subsequently increasing anterior shear force of the tibia relative to the femur, a driving mechanism for ACL injury [74,95]. Ligament strain is highest when the knee is in or near full extension (0°-30° flexion), with some studies reporting a range of 60° flexion to full extension (0°) [4,95,132]. At low flexion angles, the hamstrings are less effective in counteracting the forward pull exerted by the quadriceps. Reduced antagonist support becomes especially relevant in landing and deceleration tasks, where peak ACL forces arise from a complex interplay of anterior tibial shear force generated by quadriceps contraction and knee compression, combined with external posterior tibial shear force [74]. This interaction highlights the role of balanced agonist-antagonist activation, often assessed by the hamstring-quadriceps strength ratio [133,134]. Myer, Ford, Barber Foss, Liu, Nick and Hewett [131]prospectively found female ACL injured patients showed lower hamstring strength magnitudes than their male counterparts, yet no differences in quadricep strength.

Extrinsic Risk Factors. In addition to individual characteristics, external factors contribute to injury risk. For instance, the traction between an athlete's foot and the playing surface can affect movement control [135]. Findings from Thomson, Whiteley and Bleakley [135] systematic review revealed that higher rotational traction, increased the likelihood of ACL injury among male athletes. Higher traction combined with the rapid movements during pivoting, cutting, or decelerating, increases ligament strain, thus elevating the risk of injury. Furthermore, footwear designed has been of interest, with research showing that cleats featuring longer and more widely spaced studs around the perimeter generate greater peak rotational traction [121]. These cleats were a 3.4 greater fold of ACL injury compared to those with flat, screw-in, or pivot disk stud designs [121,135,136].

While increased traction has been associated with a heightened risk of ACL injury, research remains inconclusive on the influence of playing surface, specifically when comparing natural grass to artificial turf [121,137,138]. Grass type has been associated with a higher risk of non-contact injury in Australian male football players, particularly with Bermuda grass, compared to Rye grass [7,105]. In regards to handball court surface, no differences have been identified as a risk factor, with respect to wooden floors with lower friction, and artificial floors with higher friction [139].

Injury Mechanism

Non-contact ACL injuries generally stem from sudden deceleration, rapid pivoting, combined with high-impact forces. Cutting tasks, involving abrupt directional changes, are a common scenario due to the high demands on lower limb mechanics and neuromuscular coordination [1,21,22,73,140]. Dynamic stability and control depend on proper joint kinematic alignment and sequencing, including proper flexion-extension, rotation, and load distribution about the knee joint. Altered or improper mechanics have been associated with an increased risk of ACL injury, especially when combined with external forces 2-18 times greater than body weight, commonly seen in high-impact movements [65].

One of the key mechanisms of ACL injury is dynamic knee valgus (i.e. medial knee collapse), characterized by internal knee abduction, hip adduction, and internal hip rotation [72,73,81,141]. Examining bruised bone patterns helps to provide additional insight, revealing how the underlying forces contribute to injury [65,72,142-144]. ACL injuries affect the entire knee joint, with indirect signs of rupture frequently indicated by bruising on the lateral femoral and lateral tibial condyles [65,145]. In the instant of injury, altered lower limb alignment occurs as the knee collapses inward and directs compressive forces that bring the lateral condyles into forceful contact. Concurrently, translation and shear forces – combined with hip adduction and internal rotation – impact the medial and posterior areas of the lateral condyles. Though bruising in the lateral structures is suggested to be more common, medial compartment bruising of both the femur and tibia has also been observed in males and female patients [144,146]. Magnetic resonance imaging scans obtained one month post-ACL injury revealed femoral and tibial bone bruising patterns that are indicative of anterior tibial translation (mean 24.6-26.9 mm), internal tibial rotation (5.3°-8.6°), knee valgus (8.0°-8.3°), and slight knee flexion (20.2°-18.4°) in males and females [143]. In non-contact injuries, more severe bone bruising was indicative of greater energy transfer than contact injuries [144]. Understanding these kinematics helps us to understand further biomechanical and neuromuscular factors that contribute to ACL loading and injury.

Cutting Dynamics. As previously stated, greater risk of ACL injury is prominent in agility sports that involve frequent directional changes, producing excessive multiplanar loads at the knee joint. Retrospective studies have identified cutting as a leading action associated with non-contact injuries [5,17-19,147]. Furthermore, understanding the phases of sidestep cutting is valuable for analyzing the mechanisms that make the knee more susceptible to injury. Broadly, it can be broken down into sequence of events that follows: (1) plant and decelerate, (2) stance-phase (during this time trunk begins to reorient towards the direction you want to go), (3) swing-phase (follow through with the pivot leg prior to toe-off) [83,148]. In the occurrence of improper body alignment and stability, the ACL is put under increased strain, heightening the risk of injury.

The early percentage of the stance phase during sidestep cutting is critical for deceleration. In video analyses of injured female athletes, peak vertical GRF was observed 40 ± 0.83 ms after initial contact [149]. At that time, mean knee flexion angle was 47°, while the knee abduction angle increased by 12°, from neutral at initial contact. Additionally, mean knee internal rotation was 5° at initial contact and increased by 8° [149]. During 30-40° sidestep cutting conditions with and without a simulated defensive opponent, healthy participants exhibited peak knee valgus displacement within the first 20% of stance phase with an approach speed between 4.5 – 5.5 m/s [150,151]. Regardless of approach speed, data has shown a rapid increase in knee valgus loading occurs shortly after initial contact during cutting, indicating athletes are more susceptible to ACL injury at this time [10,141,147,151-153]. On bimodal curves, peak knee valgus angles have been identified during deceleration and sidestep phases [153,154]. In healthy female athletes performing 90° sidestep cutting tasks at a speed of 2-3 m/s, peak knee valgus angles of 11.7 ± 7.57° and 9.1 ± 3.5° were recorded with corresponding knee flexion angles of 56.3 ± 9.9° and 58.6 ± 11.7°, respectively [154]. The first peak knee valgus angle occurred during the deceleration phase, approximately 79.6 ± 38.2 ms after foot-strike, while the subsequent peak valgus angle was observed during the side movement phase, 170.6 ± 33.7 ms after foot-strike [154].

Alterations in foot placement- as the distal segment to the knee- influences joint loading as a direct consequence of the kinetic chain [150,152,155,156]. For example, lateral alignment of the foot relative to the knee joint axis increases knee valgus stress, a risk factor for ACL injury. In vivo experiments further demonstrate that internal tibial torque occurs as the body twists laterally about a planted foot [85]. Combination of valgus stress and internal tibial torque amplifies joint loading and ligament stress, elevating the potential for injury. Video analyses identified ACL injuries most often occurred when the foot was firmly planted and positioned lateral to the femur-knee axis, during deceleration of cutting [72,73,147]. In these instances, the knee was nearly extended, while positioned in valgus, and accompanied by rotation of the tibia either internally or externally [147].

Furthermore, rearfoot strike patterns have been shown to increase knee valgus and internal rotation kinematics occurrence and presence during sidestep cutting, compared to forefoot strike patterns, particularly in healthy females compared to forefoot strike [150]. Additionally, when cutting at a 45° angle with a wide foot placement of the planted limb, peak knee valgus moments reached approximately 0.79 ± 0.38 Nm/kg·m, compared to 0.45 ± 0.32 Nm/kg·m in normal stance and 0.51 ± 0.37 Nm/kg·m narrow stance sidesteps in healthy males [152]. In addition, the impact of foot progression angle has been examined on medial knee joint loading during landing and cutting tasks [155,156]. A toe-out angle (6° ± 2°) was associated with the greatest knee valgus moment (-0.82 ± 0.28 Nm/kg·m), compared to toe-in (6° ± 3°) and toe-forward (6° ± 3°) angles (-0.60 ± 0.29 Nm/kg·m and -0.67 ± 0.26 Nm/kg·m, respectively) at initial contact [156]. However, Peel, Thorsen, Schneider and Weinhandl [155] observed no significant effect of foot progression angles (15° toe-in and 15° toe-out) on internal knee adduction moment during anticipated 45° cutting tasks.

The relationship between knee valgus moments and lower extremity kinematics is critical in understanding how kinematic patterns impact knee joint loading and stability [81]. McLean, Huang and van den Bogert [157] identified a linear relationship between peak knee valgus moments and increased hip flexion, hip internal rotation, and knee abduction angles during sidestep cutting among healthy collegiate athletes. Notably, a strong correlation between hip internal rotation and peak knee valgus moments was observed for both males (r^2=0.56) and females (r^2=0.60), with females additionally having a moderate correlation between knee valgus angles and valgus moments (r^2=0.35) [157]. These findings emphasize the relationship between dynamic valgus alignment and increase valgus load, similar to a 'beam buckling mechanism' (p. 868), in which a vertical structure can suddenly bend under compressive forces, even if the external load is below the material's breaking point [157]. Furthermore, in this position, a larger moment arm for the ground reaction force is directed more medially at the knee joint, increasing valgus stress (inward joint collapse), thereby placing additional strain on surrounding tissues, such as the ACL.

Overall, existing studies have firmly identified dynamic valgus as a key injury-causing mechanism. During activities such as landing or cutting it is characterized as the inward collapse of the knee, accompanied by decreased knee flexion and increased knee abduction angles, and a greater knee adduction moment [73]. It is evident that poor neuromuscular control and inadequate joint stability contribute to injury risk. Understanding the biomechanical factors has formed the basis for injury prevention and rehabilitation. However, equally important is the consideration of an athlete's cognitive demands and its influence on performance.

Dual Tasks

Dual Tasks Overview

Interactions with the world are rarely unitarian, often requiring individuals to manage multiple sensory, cognitive, and motor demands simultaneously. Estimating how the human brain allocates cognitive attention during the implementation of cognitive-motor tasks has become a crucial area of study, shedding light on the intricate interaction between concurrent mental and physical demands. The study of dual-task performance has a rich history in cognitive and motor research [158-160]. However, the purpose of this review is to explore the biomechanical applications of dual-task conditions within these populations. Dual-task frameworks categorize and organize concurrent task performance based on factors such as task type and complexity, sensory modalities, and environmental factors [29,31,159,161]. These tasks often consist of one cognitive and one motor domain, each with distinct and separate goals [29]. Though paradigms of motor-motor or cognitive-cognitive have been presented [162], for the purposes of this review, the focus will be on cognitive-motor dual tasks, particularly relevant to the field of biomechanics.

Motor skills are acquired voluntary movements that enable purposeful movements and actions. Both physical and cognitive limitations can impair task execution, with the cognitive overload particularly affecting the ability to process and organize movement efficiently [163]. This emphasizes the importance of human cognition, which involves mental processes that help us to understand and organize the sensory input we receive from the world around us [164]. These processes include perception (acquiring and focusing attention on information), selection of attention, understanding (representation), and memory (retaining information), all of which guide behavior and action. At the core of these processes is attention, which plays a crucial role in determining how effectively one can manage and perform tasks [165,166]. Impairments in attention influence both cognitive and motor abilities, making dual-task performance more challenging [158,167].

Dual Tasks and Attention Theories

Understanding the role of attention and its impact on performance is fundamental in the context of dual tasks. Attention can be defined as both the mental effort required to focus on a task [167] and the integration and synthesis of various brain functions [165]. Task engagement can occur consciously, where actions are deliberately controlled, or non-consciously, where behaviors become automatic through repetition and practice [168]. Historically, attention was viewed as a single, unitary resource, a concept that has been widely debated [167,169,170]. As research progressed, particularly in the area of dual tasks, attention has since been recognized as a multifaceted process [160,164,165,171-173]. This understanding highlights the complexity of how attention is allocated across tasks, particularly when cognitive demands are heightened. It reflects the dynamic nature of attention, which can be divided, shifted, or alternated based on task demands [170].

Attention can be further categorized into selective, sustained, and divided, each playing a distinct role in managing cognitive resources. Selective attention refers to the allocation of cognitive resources to one task at the expense of another [170]. In such instances, individuals focus on a specific stimulus or task while filtering out irrelevant information. For example, when a defensive athlete lines up, the primary focus is on their opponent, while ignoring distractions in the environment. In contrast, divided attention is the attempt to focus on multiple tasks or stimuli at once (i.e. multitasking) [174]. However, research has shown that divided attention can impair performance in both tasks, as a result of the shared cognitive resources that are necessary for both tasks, often leading to decreased efficiency [160,175]. Conversely, alternating attention or task-switching is the ability to shift and redirect cognitive resources between tasks when multiple stimuli are involved. For instance, an athlete may alternate their attention between monitoring footwork and strategizing their game-plan while managing oncoming defense. This type of attention is believed to increase cognitive load, which can negatively impact performance [176,177]. Sensory-specific attention systems, such as auditory, visual, and proprioceptive attention play a critical role in biomechanics and task performance by guiding motor actions. Together, these sensory systems provide necessary feedback to regulate and refine motor actions. These modalities will be discussed within the context of attention theories in the following sections.

Dual-task procedures are used to assess attention demands of performing a motor-skill. Performance of one task is often sought to be “easy”, dependent on the performers’ skill level;

however, execution of two concurrent tasks is increasingly difficult. Theoretical perspectives help to provide an understanding of attention limitations. These modalities will be discussed in the context of two broad categories: bottleneck (filter) theories and capacity sharing models [167,170,178].

Bottleneck Theory. Bottleneck theories suggest that neural processors and networks are incapable of multitasking. Notably, divided attention is rebutted by the brains' inherent nature to serial process signals [179,180]. Theory developments have suggested that information not selected for further processing is filtered out at various stages of the cognitive process [180,181]. Initial framework posited that the bottleneck occurred at the early stages of information processing, particularly sensory registration, and detection. Others have argued that filtering happens later, just prior to the response selection stage [182]. When multiple stimuli are presented, the brain can only attend to one at a time, delaying the processing of other stimuli [179,180], or leaving them not responded to altogether [167]. Bottleneck theory assumptions are often violated, due to the need to establish a response order, which results in delayed processing [34]. Moreover, the evolution of bottleneck theories fails to fully account for the flexibility and adaptability of human cognition [167,183]. Concurrent and complex tasks often involve multiple stimuli, and while bottleneck theories assume strict processing limitations, capacity models of attention offer a more adaptive explanation for how the brain manages attention across multiple tasks, suggesting that cognitive resources can be dynamically allocated.

Capacity Model Theory of Attention. Alternative attention theories, such as the central-resource theory, propose that attention (i.e. cognitive effort) has a finite capacity [167,184,185]. This model argues that attention for all activities-whether cognitive or motor- compete for mental resources from a single pool [167,171,186]. Kahneman's model of attention (1973) illustrates how cognitive resources are allocated, with incoming stimuli being recognized and categorized. The amount of attention described as cognitive effort is related to the resources required to execute the tasks at hand, and largely dependent on task complexity.

Distribution of cognitive effort (i.e., resources) is influenced by factors such as arousal level (cognitive excitability), individual dispositions, task requirements (i.e., novelty), and environmental context. Along with these, the mental resources required to manage and process information directly impact performance outcomes [167]. While this theory suggests attention can be flexibly allocated across tasks, a notable limited pool of resources is considered. Both cognitive and motor tasks must compete for these finite attention resources, corroborating why performance may decline under dual-task conditions. Moreover, a limitation of the central-resource theory is its difficulty in accounting for how individuals can perform different types of tasks simultaneously, particularly when those rely on different modalities or sensory systems. Building upon the example above, driving a car while engaging in a conversation requires both visual-spatial and auditory processing [168]. Despite the tasks involving different sensory systems, performance in one or both tasks may still decline, challenging the idea of a single, undifferentiated pool. This theory falls short in explaining how individuals can manage tasks involving various sensory modalities.

To provide a more comprehensive explanation, multiple resource theory was introduced, offering a more nuanced perspective [175,187,188]. It proposed cognitive resources are not drawn from a single pool, but from multiple independent pools based on sensory modality, processing stage, and required task response [175,185]. Under this framework, task type greatly influence the source of information processing, and can be categorized into three areas: 1) input and output modalities, such as vision, limbs, and speech systems; 2) stages of information processing including perception, memory encoding, and response output; and 3) codes of processing information, such as verbal and spatial codes [168]. This framework assumes that sensory modalities draw from separate resource pools. However, theoretical implications arise when the demands of a task exceed the available capacity, depleting the remaining pools and consequently affecting task performance [178]. Furthermore, it does not account for task novelty respective to the performer.

Evolution of attention theories has led to the consensus that attention is not a single entity or mechanism, but rather a collection of various biological processes for cognitive processing [170]. In practical situations, task demands often overlap, leading to bottlenecks [160,183,189]. When multiple tasks compete for limited cognitive resources, performance can be delayed or degraded [171]. Despite existence of multiple resource pools for various tasks, cognitive capacity depletes with increased task complexity, diminishing the efficiency of executive control mechanisms [181]. Furthermore, functional magnetic resonance image studies during dual tasks have shown that there is no distinct ‘multitasking region’ in the brain responsible for processing tasks simultaneously, suggesting that the brain relies on distributed networks for handling concurrent tasks [189]. Brain activity showed increased interference with greater overlap in sensory stimuli, as well as when more cognitive resources were required due to increased complexity [189]. These results demonstrate that the brain relies on a network to manage concurrent tasks, with interference being an inherent consequence when cognitive resources are overloaded.

Dual Task Interference

Building on the attention theories presented, we can better understand how cognitive demands lead to interference in performance. Concurrent tasks that rely on similar perceptual processes can increase interference by competing for the same cognitive resources [173,185,190]. Moreover, the demands of concurrent tasks inherently increase cognitive complexity, which can further hinder performance compared to when each task is performed separately [190]. Interference mechanisms can also stem from function- and data-specific limitations [190]. Data-specific (i.e., cognitive) limitations occur when tasks process similar information. For example, athletes often need to respond to simultaneous visual stimuli (e.g., visual cues indicating the direction of the cut) and auditory stimuli (e.g., a coach's verbal command). Function-specific limitations refer to tasks that typically share similar motor processes, such as dribbling a ball and cutting. When these tasks are performed simultaneously, competition for motor resources (e.g., coordination, proprioception, and stability) may deter performance or increase injury risk [15]. Understanding the mechanisms that influence interference enables the examination of how it manifests among various tasks.

Patterns of Interference. Cognitive motor interference is dependent on several factors such as task type, difficulty level, prioritization instructions, and performer characteristics [39,160,186,191]. Plummer, Eskes, Wallace, Giuffrida, Fraas, Campbell, Clifton and Skidmore [32] proposed a framework for categorizing patterns of cognitive-motor interference (CMI) in cognitive-motor tasks, relative to the independent performances in single-task conditions. The framework identifies interference patterns through nine potential outcomes: "1.) *no interference*- performance of either task does not change relative to *single-task performance*, 2.) *cognitive-*

related motor interference- cognitive performance remains stable while motor performance deteriorates, 3.) *motor-related cognitive interference*- motor performance remains stable while cognitive performance deteriorates, 4.) *motor facilitation*- cognitive performance remains stable while motor performance improves, 5.) *cognitive facilitation*- motor performance remains stable while cognitive performance improves, 6.) *cognitive-priority trade off*- cognitive performance improves while motor performance deteriorates, 7.) *motor-priority trade off*- motor performance improves while cognitive performance deteriorates, 8.) *mutual interference*- performance of both tasks deteriorates, and 9.) *mutual facilitation*- performance of both tasks improve" (p. 2). While some patterns are observed more frequently than others, there is limited application of these findings in the context of ACL research.

In biomechanics research, dual task paradigms are used to understand how cognitive demands impact motor performance. Despite existing research incorporating cognitive tasks with movements such as cutting and landing, current assessment methods lack robustness. In several cases, researchers categorized participants as high or low performers based on cognitive assessment scores, then examine biomechanical outcomes during landing movements based on these classifications [13,15,37]. Avedesian, Forbes, Covassin and Dufek [14] systematic review found that participants with lower cognitive performance were associated with higher musculoskeletal injury risks. However, in these cases, researchers have failed to capture how the interaction between cognitive load and motor task performance causes shifts in attention, which in turn can affect performance outcomes.

In contrast to previous studies linking low-cognitive groups with greater injury risk mechanics, Fischer, Hutchison, Becker and Monfort [192] assessed participant outcomes during dual task conditions involving both anticipated and unanticipated jump-landing scenarios. Participants exhibited cognitive-motor interference, showing reduced peak knee flexion angles during dual tasks compared to baseline trials. These dual tasks challenged working memory, decision making, and controlled processing, making it a more comprehensive protocol that allows for a deeper understanding of interference [31,192]. Furthermore, to assess the patterns of interference effectively, it is crucial to understand robust methods that can evaluate the nuanced interactions between cognitive load and motor performance.

Assessing Interference in Performance. Interference can be assessed through performance-based assessments, which compare the outcomes of dual task performance to baseline single task measurements, and cognitive assessments, which evaluate the cognitive load and processing efficiency, often using metrics like reaction time in the psychological refractory period (PRP) effect. When tasks are presented in close succession, decreased performance- often observed as a slower response or execution time in the subsequent task- is known as the psychological refractory period (PRP) effect [180,193]. As the time gap between two tasks- known as the stimulus onset asynchrony (SOA)- decreases, PRP becomes more pronounced [180]. This was observed when participants were instructed to press a button in response to a light cue when walking. Longer reaction times occurred when participants had to walk and cross an obstacle with high visual sensory interference compared to low sensory interference ($p<0.001$) [194].

Attentional demands of dual tasks refer to the cognitive resources required to manage and allocate resources tasks. Cognitive motor interference (CMI) is the measure of deterioration in performance of one or both tasks, often observed as delayed response times, or decreases in accuracy [32,195]. To quantify interference, baseline single-task performance is measured and compared to dual task outcomes Kelly, Janke and Shumway-Cook [196] measured absolute and relative performance changes, known as dual task cost (i.e., effect) (DTC): $\frac{(dual\ task - signle\ task)}{single\ task} \times 100\%$. Negative and positive DTC values indicate performance decrement (dual task cost) and improvement (dual task benefit), respectively. In the presence of response decrement, response latency was calculated $\frac{-(dual\ task - signle\ task)}{single\ task} \times 100\%$.

Attentional trade-offs within tasks are measured using the attention allocation index (AAI), which measures how attention is distributed between motor and cognitive tasks. This can be calculated as: $\frac{dual\ task\ motor\ performance - single\ task\ motor\ performance}{dual\ task\ cognitive\ performance - single\ task\ cognitive\ performance}$ [197]. A positive AAI indicates motor performance is favored, whereas a negative value suggests that cognitive performance is prioritized. Kelly, Janke and Shumway-Cook [196] measured task trade-offs during walking trials where participants simultaneously performed the Stroop test. The difference between the walking-focused DTC and the cognitive-focused DTC was used to assess changes within the walking variables. Conversely, the difference between the cognitive-focused DTC and walking-focused DTC assessed changes in the cognitive variables. When participants were instructed to focus on the walking task, response latency increased (0.83 s) compared to when participants focused on the cognitive task (0.69s) [196]. These results demonstrate the cognitive demand (i.e., effort) required for various tasks.

To further evaluate the relationship between task outcomes, Performance Operating Characteristic curves are implemented (POC) [178]. This graphical representation illustrates the trade-offs between two concurrent tasks, allowing researchers to visualize how the performance of one task varies in relation to the other. The POC helps to determine the degree of interference between the tasks and how task difficulty, attentional focus, or external factors impact overall performance. Plotting the DTCs for both cognitive and motor tasks provide visual observation of any shifts in attention and performance [198].

By quantifying dual task costs and effects, and examining attentional trade-offs through AAI and POC curves, valuable insight is gained into how attention is distributed between concurrent tasks. Furthermore, these methods help reveal the underlying patterns of interference that emerge when cognitive and motor tasks compete for resources.

Dual Tasks Taxonomies and Applications

Understanding interference patterns is critical for applying dual-task taxonomy in biomechanics settings. Across the range of dual-task frameworks, the required concurrent motor and cognitive tasks must have two separate and distinct goals [29]. Gentile's taxonomy of motor skills established a system to categorize movement tasks based on environmental context and action functions [163]. McIsaac, Lamberg and Muratori [29] extended upon Gentile's taxonomy to establish a physician's perspective on categorizing and standardizing dual task paradigms into three categories: action goal, complexity, and novelty. Task complexity refers to the cognitive demands of attentional and cognitive resources. In alignment with the Challenge Point Framework [199], "nominal difficulty"- the inherent difficulty level, regardless of the performer's skill level- helps describe the baseline challenge of a task. "Functional task difficulty" accounts for the performer's skill level relative to the task. In McIsaac's proposed framework, task classification varies along a continuum of low and high novelty and complexity [29].

Dual-task testing methods revealed performance decrease as task complexity increased in experienced professional rugby players [35]. Additionally, athletes demonstrated a significant correlation (r^2=0.52) between performance and task complexity, indicating that players who excelled in the simple task also tended to perform well in the more challenging task [35]. These results highlight the importance of incorporating standardized measures of task complexity in dual task testing to ensure consistency in performance assessment.

Moreover, Wollesen, Wanstrath, van Schooten and Delbaere [31] categorized task types based on execution, such as reaction time, controlled processing, visuospatial, mental tracking, working memory, and discrimination tasks [200]. The following examples demonstrate how various dual-task paradigms are applied in biomechanics literature. However, it is important to note that dual-task implementation extends beyond these. For more comprehensive reviews pertaining to gait in healthy individuals, refer to Al-Yahya, Dawes, Smith, Dennis, Howells and Cockburn [201], Bayot, Dujardin, Tard, Defebvre, Bonnet, Allart and Delval [34] and Wollesen, Wanstrath, van Schooten and Delbaere [31]. Clinical neurological applications are discussed in Plummer, Eskes, Wallace, Giuffrida, Fraas, Campbell, Clifton and Skidmore [32], while sports-related applications are reviewed by Chaput, Ness, Lucas and Zimney [20], [202], [161], and [203].

Choice-reaction tasks estimate cognitive processing speeds, often by measuring how quickly an individual responds to a stimulus. In a choice-reaction dual task, participants were required to respond to a light stimulus while performing a single leg land [2]. Dual task trials exhibited higher peak vertical ground reaction forces (GRF) compared to catch trials- when participants were instructed to react to the stimulus, yet no light was presented [2]. No significant differences among reaction times were observed, indicating cognitive-related motor interference [2]. Similar findings were observed in controlled processing tasks, where participants not only had to respond to a stimulus but also engage in a decision-making process [3]. Twenty uninjured females (18-25 years) exhibited larger peak vertical GRF magnitudes (1.75 ± 0.51 BW) when they were required to reach overhead and grab a ball suspended above the force plates, while simultaneously responding to a stimulus that determined landing technique, compared to the control condition (1.62 ± 0.38 BW) [3]. Controlled processing tasks involve processing speeds, and deliberate, conscious attention for decision making. While choice-reaction tasks rely heavily on automatic responses to external stimuli, more complex cognitive demands that require multiple steps are classified as controlled processing.

Furthermore, in visuospatial tasks, where individuals must actively interpret and manipulate visual and spatial information, participants may be required to remember the position of an object. For example, variations of the Clock Test have been implemented during cognitive-motor walking conditions, where the participant verbally determined whether or not the minute and hour hands would be on the same side of the clock, in response to being given a time of day [204-206]. An alternative visual memory task required participants to recognize and remember what light stimulus corresponded with "go" during triple hop tests [207]. During the visual-cognitive reaction conditions, maximum hop distance significantly decreased by 8.17% compared to control conditions [207]. Visuospatial tasks demand higher cognitive processing skills such as memory, attention, and problem-solving. With regards to shifting the focus on maintaining and updating information over time, mental tracking is a common task modality implemented in biomechanics dual-task literature [31].

Mental tracking tasks involve holding information while actively engaging in a cognitive process. Common examples are arithmetic and verbal fluency tasks [31,32]. In a study on backwards serial counting by 3s and by 7s, cognitive complexity declined with task complexity in both healthy and ACL-injured male adults while walking [208]. For the easier task (counting by 3s), cognitive scores decreased during normal walking (46.3 ± 21.0) and narrow walking (43.4 ± 17.9) compared to baseline measurements (49.3 ± 21.7) in the ACL-injured group. When cognitive task complexity increased (counting by 7s), scores further decreased in normal (25.4 ± 12.9) and narrow (24.0 ± 13.5) walking compared to baseline scores (27.0 ± 13.2) [208]. Gait parameters of step length and step velocity remained unchanged, showing motor-related cognitive interference [208]. On the other hand, Shi, Huang, Yu, Liang, Zhang, Yu, Liu and Ao [209] reported reduced gait speeds during self-selected walking trials while backwards counting by 7s in ACL patients. However, no cognitive task scores were reported, to further explore the interference pattern. Related to mental tracking tracks, yet distinct, is working memory.

While mental tracking requires holding information and manipulation, working memory tasks only require holding information for a short period of time [210,211]. Working memory load was manipulated using an n-back task while older adult participants walked on a treadmill [212]. However, no dual-task effects were observed, indicating no cognitive interference [212]. In contrast, perceptual-cognitive training, a type of training to improve response to sensory information in complex environments, exhibited effective cognitive results in volleyball players during an 8-week intervention [213]. Though Fleddermann, Heppe and Zentgraf [213] results suggests cognitive capacity is malleable, and can improve with training, their findings also highlight limitations in dual-task performance. Specifically, during cognitive-motor tasks, participants demonstrated a decline in motor performance [213]. The observed cognitive priority trade-off interference emphasizes the limitations of working memory capacity in mental tracking tasks. Similarly, in discrimination tasks, working memory is crucial for filtering out relevant information, but a key distinction is that individuals must also respond accordingly.

Discrimination tasks involve identifying important details and distinguishing between stimuli, requiring individuals to make accurate responses based on those distinctions. In a study by Pitman, Sutherland and Vallis [214], participants walked a 7-meter path with an obstacle in the way, while responding to the cognitive Stroop test [214]. Visual cues (LED lights) governed participants to either step over or circumvent the obstacle. Successful task performance required participants to simultaneously manage the Stroop test and obstacle navigation. As the visuomotor task complexity increased, participants reduced walking speed by an average of 0.05±0.01 m/s, indicating cognitive-related motor interference [214].

Overall, this section has reviewed the types of tasks implemented in concurrent cognitive-motor paradigms. Although not exhaustive, the findings discussed highlight the variability in performance outcomes and interference patterns across different task types. Classification of the secondary cognitive task, regard it to be *incorporated* into the motor task or the motor task is completed with an *additional* (i.e. distractor) cognitive task [162]. Implementation of an *additional* task is unrelated to the primary task [162,215]. This is seen in studies when participants are required to solve arithmetic problems during gait trials. For clinical applications, Chaaban, Turner and Padua [202] advises the secondary cognitive tasks must be manageable alongside the motor task. While also incorporating quantifiable assessments (i.e. reaction time or accuracy), and ensure consistent, reliable outcomes suitable for repeated use [202].

Not only are performance outcomes highly influenced by task type but also depend on performers' differences in skill level. For example, choice-reaction tasks are frequently used in biomechanics research to assess injury risk, such as in cutting tasks designed as anticipated or unanticipated are often implemented in injury risk assessment studies in biomechanics research, in the form of cutting tasks, designed as anticipated and unanticipated [15,216,217]. However, these methodologies often fall short of true dual-task requirements due to the absence of an additional cognitive component. To improve these assessments, incorporating tasks that impose greater cognitive demands alongside motor tasks, both having two separate and distinct goals, is necessary to better assess attentional demands.

Conclusion

Understanding that interactions with the world are rarely unitarian posits that non-contact ACL injuries are influenced not only by mechanics but also cognitive factors. Estimating how the human brain allocates cognitive attention during the implementation of cognitive-motor tasks has become a crucial area of study, shedding light on the intricate interaction between concurrent mental and physical demands. The study of dual-task performance has a rich history in cognitive and motor research [158-160], and become increasingly relevant among athletes recovering from ACL injuries, where cognitive-motor interactions play a key role in regaining functional abilities [16,20,162,218,219]. However, few studies exist that have explored the interaction between cognitive-motor dual tasks to explain interference. This information can better address the cognitive and motor challenges encountered by athletic populations, particularly those at greater risk for injury, and ultimately improve performance outcomes and rehabilitation research.

Chapter 3: Methods

Participants

Eighteen participants were recruited; however, data from one participant were excluded due to data collection error, resulting in a final sample of seventeen healthy females (22.4 ± 2.8 years; height: 1.69 ± 0.05 m; body mass: 62.9 ± 7.5 kg) participated in this study. A sensitivity analysis was conducted in G*Power (Version 3.1; Faul et al., 2007) for a 2 × 2 within-subject design to estimate the minimum detectable effect size for the study sample. The analysis specified an alpha level of 0.05, statistical power of 0.80, and a correlation of 0.60 among the four within-subject conditions (COG_AN, COG_UN, NOCOG_AN, NOCOG_UN), assuming sphericity ($\varepsilon = 1$). With a total sample of 18 participants, the design was sufficiently powered to detect effects of $f = 0.26$ (equivalent to $\eta p^2 \approx 0.07$), representing a medium effect size.

Prior to data collection, ethical approval from the university's Institutional Review Board was received. Inclusion criteria was as follows: females aged 18 to 30 years, with no history of lower extremity injury in the six months prior to testing, no history of lower extremity surgeries, and recreationally active—defined as engaging in moderate-intensity aerobic activity at least five days per week for a minimum of 30 minutes per session, or vigorous intensity aerobic activity three days per week for a minimum of 20 minutes per session. Prior to data collection, participants completed a musculoskeletal health history questionnaire and a fitness history questionnaire. Participants also reported on previous sport experience. Participants were asked to wear an athletic sports bra or tank top, spandex shorts, and the provided laboratory shoes (Nike Pegasus, Nike Inc., Beaverton, OR USA) during testing.

Instrumentation

A Trigno™ Wireless EMG system (2000 Hz, Delsys, Inc., Natick, MA, USA) measure muscle activity during the cutting tasks. MVIC testing was conducted using a dynamometer (Biodex Medical Systems, System 4 Pro, NY, USA). Raw marker kinematic data were collected using a 12-camera motion capture system (200 Hz, Vicon, Centennial, CO, USA). Synchronous ground reaction force data were collected via AMTI force plates (2000 Hz, AMTI, Inc., Watertown, MA, USA). LabVIEW VI was used to present the visual stimulus on a computer monitor placed 3 m from the force plate. The Polar Verity Sense optical heart rate sensor measured heart rate (Polar Electro, USA). Statistical analysis was completed in Python (v3.11.10) using custom scripts and the built in libraries.

Experimental Protocol

First, participants completed baseline digit recall testing under controlled conditions without any physical activity. During baseline testing, participants were shown twelve sequences of five consecutive digits and asked to recall them verbally, only when prompted by the researcher. Pauses of 10 to 30 seconds were provided between sequences to replicate the memory retention period that would occur during movement trials. Responses were recorded by the researcher on a standard data collection sheet.

Participant's height and weight was measured via a stadiometer and force plate, respectively. Electromyography (EMG) electrodes were placed in accordance with the SENIAM Guidelines [220] on the right and left vastus medialis and medial hamstring. Skin preparation involved shaving the hair, abrading the skin, and cleaning with isopropyl alcohol. Once the sensors were placed, participants completed a three-minute warm-up at a self-selected pace on a treadmill. Participants then completed maximum voluntary isokinetic testing (MVIC) for each EMG sensor while seated on the Biodex dynamometer with their hips flexed at 90 degrees. The first repetition of each served as practice, followed by two maximum effort repetitions conducted at 30 degrees per second for concentric quadriceps and hamstring contractions. Following this, a heart rate monitor was secured to the participants left arm (Polar FT1 Heart Rate Monitor, Polar Electro Inc., Bethpage, NY, USA).

Twenty-seven retro-reflective markers were placed on the manubrium below the sternal notch, C7, L1, and bilaterally on the right and left acromion processes, iliac crests, anterior superior iliac spines (ASIS), posterior superior iliac spines (PSIS), greater trochanters medial and lateral femoral epicondyles, medial and lateral malleoli, tips of the second toes, and the first and fifth metatarsophalangeal joints. Clusters of four non-collinear markers on rigid thermoplastic shells were secured using FlexiTape to the posterior trunk, posterior pelvis, and bilaterally to the lateral thighs and shanks, and to the posterior surface of both shoe heels. Once a standing static calibration trial was recorded, the 27 anatomical markers were removed.

Participants completed trials of 45-degree run-and-cut under four conditions: 1) anticipated with no cognitive task (NOCOG_AN), 2) unanticipated with no cognitive task (NOCOG_UN), 3) anticipated with a cognitive task (COG_AN), and 4) unanticipated with a cognitive task (COG_UN). To control order effects, the cognitive load conditions (cognitive vs. no cognitive) were counterbalanced across participants, while the anticipation conditions (anticipated vs unanticipated) were randomized. Participants first completed practice trials at a self-selected 'fast, game-like' pace. Practice continued until mean approach and exit speeds met the 3.25 m/s threshold and demonstrated consistency, defined as a coefficient of variation within

±2 SD of each participant's preferred speed. A trial was considered successful if the participant planted their full foot on the force plate, executed the correct cut direction, avoided cross-over steps, and achieved the minimum approach and exit speeds of 3.25 m/s. Trials were re-randomized and repeated if success criteria were not met or if EMG or motion-capture data were missing or corrupted due to equipment error. Each participant completed six successful trials per condition (three to each side).

For the trials with an added cognitive task, participants performed a working-memory task while executing a 45-degree cut [161]. Participants were shown 3 consecutive digits at 1 second per digit, just before they start their run and cut task. After completion of the cut, participants were asked to verbally recall the digit sequence. The researcher recorded responses in real-time using a standardized data collection sheet.

To monitor fatigue and prevent overexertion, participants were asked their heart rate and self-reported rate of perceived exertion using a 1-10 scale (10 represents maximal exertion) every few trials. Between every trial sufficient rest time was provided, as determined by the participant.

Data Reduction and Analysis

Visual3D biomechanical software (v2025.01.2, C-Motion, Germantown, MD, USA) was used to process and analyze three-dimensional hip, knee, and ankle joint kinematic and kinetic variables. Raw marker coordinate and ground reaction force data were filtered using a fourth-order, zero lag, Butterworth low-pass filter with a cut-off frequency of 12 Hz [97,221]. Three-dimensional angular computations were performed using Right-handed Cartesian segmental coordinate systems (x-y-z), where the x-axis represents the mediolateral direction, y represents anteroposterior, and z represents the longitudinal axis. Hip joint centers were defined at 25% of the distance from the ipsilateral to contralateral greater trochanter markers [222], knee joint centers were determined using the midpoint between femoral epicondyle markers [223], and ankle joint centers were determined using the midpoint between medial and lateral ipsilateral markers [224].

Raw EMG and MVIC data were processed using a fourth-order zero-lag order high pass Butterworth filter with a cutoff frequency of 10 Hz, followed by full wave rectification. The rectified signal was then low-pass filtered at 6 Hz to obtain the linear envelope[225]. MVIC EMG signals were smoothed with a 250-sample moving average filter (~125 ms at 2000 Hz) to reduce transient fluctuations and stabilize burst amplitude estimates. For each MVIC trial of consecutive isokinetic knee flexion/extension repetitions, the largest two amplitude bursts were identified, and the maximum amplitude within each burst was extracted. The average of these two maxima was taken as the peak MVIC value for normalizing the movement trials EMG data.

EMG trials were trimmed from 50 ms before to 100 ms after initial force plate contact. Mean normalized EMG amplitude was calculated within two windows: pre-contact (-50 ms to initial contact) and stance (0-100 ms post initial contact). Quadricep-to-hamstring (Q/H) co-activation ratio was also computed for both windows, defined as

$$Q/H_{ratio} = \frac{mean\ vastus\ medialis\ activity}{mean\ medial\ hamstring\ activity}.$$

A Q/H ratio greater than one indicates greater quadriceps relative to hamstrings activity, whereas values less than one indicate greater hamstrings relative to quadriceps activity.

Recalled digit sequences were compared to the original sequence presented. Responses were scored based on the number of digits recalled accurately in the correct order. Each correctly recalled digit in its correct position received one point. The total score for each trial was the sum of the correctly recalled digits. Digit recall scores from the cognitive run-and-cut tasks were compared to the baseline test scores. Baseline performance was defined as the average across all twelve baseline test trials. For each participant, dual task cost (DTC) was calculated separately for anticipated and unanticipated conditions in two ways: (1) as the raw difference between baseline and dual task performance:

$$DTC = Performance_{dual\ task} - Performance_{baseline},$$

and (2) as the percentage change from baseline, using the formula:

$$DTC\% = \frac{(Performance_{dual\ task} - Performance_{baseline})}{Performance_{baseline}} \times 100.$$

Positive DTC values indicate poorer performance under dual task conditions relative to baseline whereas negative values indicate improved performance under dual task conditions.

Statistical Analysis

All data processing and analyses were completed in Python (v3.11.10) using custom scripts and the built in libraries: NumPy, pandas, SciPy, statsmodels, and pingouin. Linear mixed models (LMMs) were implemented to examine main and interaction effects of anticipation and cognitive load on each kinematic, kinetic and EMG variable. Each model followed a 2×2 factorial structure with *Anticipation* (Anticipated, Unanticipated) and *Cognitive Load* (Cognitive, No-Cognitive) as fixed factors, and a random intercept for *Participant* to account for within-subject dependence:

$$Y_{ijk} = \beta_0 + \beta_1(Anticipation_j) + \beta_2(CognitiveLoad_k) + \beta_3(Anticipation_j \times CognitiveLoad_k) + u_{0i} + \varepsilon_{ijk}$$

where:

- Y_{ijk} = dependent kinematic outcome for participant *i* under anticipation level *j* and cognitive load level *k*,
- β_0 = model intercept (grand mean),
- $\beta_1, \beta_2, \beta_3$ = fixed-effect coefficients for anticipation, cognitive load, and their interaction,
- u_{0i} = random intercept for participant *i*
- ε_{ijk}= residual error term, assumed ε_{ijk}~N(0,σ^2)

In the linear mixed model, the main effect reflects the difference between two conditions, combining across the other factor (anticipation and cognitive load). For example, the anticipation main effect takes the average dependent variable value from both unanticipated trials and compares it to the average from both anticipated trials:

Anticipation main effect:

$$= ((COG_{unanticipated} + NOCOG_{unanticipated})/2) - ((COG_{anticipated} + NOCOG_{anticipated})/2)$$

Cognitive load main effect:

$$= ((COG_{Anticipated} + COG_{Unanticipated})/2) - ((NOCOG_{Anticipated} + NOCOG_{Unanticipated})/2)$$

Interaction Term:

$$= \left(COG_{Unanticipaated} - NOCOG_{Unanticipaated}\right) - \left(COG_{Anticipaated} - NOCOG_{Anticipaated}\right)$$

Models were estimated using maximum likelihood (ML) estimation. Fixed effect coefficients (β), standard errors, t-statistics, and p-values were extracted from model summaries. Post-hoc pairwise contrasts and estimated marginal means (EMMs) were computed using pingouin**'s** emmeans and pairwise functions, with Holm–Bonferroni correction for multiple comparisons. Model assumptions, including normality of residuals and homoscedasticity, were assessed using visual inspection of Q-Q plots and residual plots.

Chapter 4: Main Effects and Interaction Effects of Working Memory on Lower Extremity Biomechanics and Muscle Activity

Abstract

Anterior cruciate ligaments (ACL) injuries are common with unanticipated cutting maneuvers, yet the influence and interaction of anticipation on lower extremity mechanics during these tasks remain unclear. This study examined the effects of anticipation and working-memory load on knee biomechanics during 45° sidestep cutting. Seventeen physically active females (22.35 ± 2.78 yrs). Performed anticipated and unanticipated cutting trials under cognitive and no-cognitive conditions. Knee joint kinematics and kinetics were collected with three-dimensional motion capture and force plates. Surface EMG was recorded from the vastus lateralis and semimembranosus. Linear mixed-effects models tested for main and interaction effects (alpha = 0.05).

A significant anticipation x cognitive interaction was observed for posterior GRF. Anticipated trials generally produced greater posterior forces; however, under cognitive load, unanticipated trials showed a relatively larger posterior magnitude. In addition, anticipated conditions exhibited larger peak knee extensor moments. In contrast, unanticipated trials showed reduced braking forces and greater knee flexion at initial contact and peak, suggesting load was managed through increased joint excursion. Under cognitive load, the quadricep-hamstring ratio was balanced pre-contact (0.96) and reduced during stance (2.60 vs 3.67). Together, these findings highlight how anticipation and cognitive load modify knee joint mechanics, particularly in the sagittal plane, and relative to ACL injury risk. These approaches may improve ecological validity and inform return-to-play programs that address both motor and cognitive factors influencing ACL injury risk.

Introduction

The anterior cruciate ligament (ACL) is an essential structure for maintaining knee joint stability. However, ACL injuries remain highly prevalent, with an estimated 200,000 injuries occurring annually in the United States [54], nearly 70-80% of which are non-contact[4,5,108]. Extensive research has established a range of risk factors contributing to injury, including both intrinsic (e.g., anatomical features, hormone profiles, and neuromuscular properties) and extrinsic risk factors (e.g., playing surface, footwear, and sport-specific demands). These factors heighten susceptibility to injury when combined with the high physical demands of sport, where non-contact injuries frequently occur during rapid deceleration and change-of-direction movements such as cutting or pivoting. Such injury mechanisms are characterized by decreased knee flexion angle, increased knee abduction angle, and increased internal knee adduction moment [22]. Coupled with altered joint mechanics, elevated ground reaction forces further increase ligament strain. In addition to these mechanical factors, neuromuscular function plays a critical role in dynamic knee stability, such as the quadriceps-to-hamstrings co-activation patterns. Pertaining to ACL injury, relative increased quadriceps activity has been identified as contributors to increased anterior tibial translation and valgus knee loading thereby elevating risk of injury. However, how these neuromuscular strategies adapt under dual task conditions remains underexplored. Moreover, despite the extensive understanding of these mechanical factors, they capture only part of the context in which injuries occur.

The nature of these movements in sport typically includes cognitive demands, such as decision making under uncertainty, or recalling a play. In these situations, athletes must simultaneously process information, select responses, and execute motor actions. This overlap suggests that concurrent cognitive-motor demands, particularly under unanticipated conditions, may alter lower extremity biomechanics in ways that are not yet fully understood. Among cognitive processes, working memory is highly relevant, as athletes must retain and update task-related information (e.g., play strategies, opponent positioning) while executing proper movement dynamics, thus, making it a critical factor for investigation in this context. Prior dual-task studies have shown that cognitive load can modify gait mechanics, balance, and postural control [34,201], yet little is known about how these effects extend to sport contexts. This is particularly important when athletes perform rapid cutting, landing, and agility tasks under cognitive demand—situations that inherently carry elevated ACL injury risk [226]. Together, these considerations underscore the importance of investigating the role of working memory in dual-task sport movements, such as cutting.

Thus, the purpose of this study was to examine the interaction between anticipation and increased cognitive load by way of a digit recall task on knee joint biomechanics in run-and-cut conditions. It was hypothesized that the cognitive-motor dual task would elicit greater ACL injury-related mechanics. Specifically, trials performed with the working memory task were expected to demonstrate decreased knee flexion angles and increased knee abduction angles, internal knee adduction moments, and peak vertical ground reaction force. As a secondary aim, EMG measures were included to provide additional insight into possible neuromuscular strategies adopted under dual task conditions. It was expected that quadricep and hamstring activation would decrease under working memory conditions, yet with a shift toward greater quadriceps dominance reflected in quadricep-to-hamstring ratios.

Methods

Participants

Eighteen participants were recruited; however, data from one participant were excluded due to data collection error, resulting in a final sample of seventeen healthy females (22.35 ± 2.78 years; height: 1.69 ± 0.05 m; body mass: 62.96 ± 7.54 kg) reported no history of lower extremity injury within the past six months and no prior lower extremity surgery. Inclusion criteria required that participants engaged in regular physical activity, defined as meeting American College of sports Medicine guidelines of at least 150 minutes of moderate-intensity or 75 minutes of vigorous intensity activity per week. Prior to data collection, experimental procedures received ethical approval from the university's Institutional Review Board and written informed consent was obtained.

Instrumentation

A Trigno^{TM} Wireless EMG system (2000 Hz, Delsys, Inc., Natick, MA, USA) measured muscle activity during the cutting tasks. MVIC testing was conducted using a dynamometer (Biodex Medical Systems, System 4 Pro, NY, USA). Raw marker kinematic data will be collected using a 12-camera motion capture system (200 Hz, Vicon, Centennial, CO, USA). Twenty-seven retro-reflective markers were placed on the manubrium below the sternal notch, C7, L1, and bilaterally on the right and left acromion processes, iliac crests, anterior superior iliac spines (ASIS), posterior superior iliac spines (PSIS), greater trochanters medial and lateral femoral epicondyles, medial and lateral malleoli, tips of the second toes, and the first and fifth metatarsophalangeal joints. Clusters of four non-collinear markers on rigid thermoplastic shells were secured using neoprene straps to the posterior trunk, posterior pelvis, and bilaterally to the lateral thighs and shanks, and to the posterior surface of both shoe heels. Once a standing static calibration trial is recorded, the 27 anatomical markers were removed. Synchronous ground reaction force data were collected via AMTI force plates (2000 Hz, AMTI, Inc., Watertown, MA, USA). LabVIEW VI was used to present the visual stimulus on a computer monitor placed 3 m from the force plate. The Polar Verity Sense optical heart rate sensor measured heart rate (Polar Electro, USA). Statistical analysis was completed in SPSS (IBM SPSS Statistics, Version 29, Armonk, NY, USA) and Python (3.10, Python Software Foundation, 2024).

Experimental Procedures

First, participants completed a baseline digit recall test while sitting in front of a television screen. The baseline cognitive test consisted of twelve unique sequences of five consecutive digits on the screen – the same screen that was later used for the movement trials. Each sequence was presented one digit at a time with each digit displayed on a separate slide of a PowerPoint presentation. Blank screens were used to separate the sequences. Participants were instructed to recall the sequences verbally, only when prompted by the researcher. Pauses of 10 to 30 seconds were provided between sequences to replicate the memory retention period that would occur during movement trials. Responses were recorded by the researcher on a standardized data collection sheet.

Next, electromyography (EMG) sensors were placed on the right and left vastus medialis and medial hamstring in accordance with the SENIAM guidelines [220]. Participants then completed a 3-minute treadmill warm-up at a self-selected pace, followed by maximal voluntary isokinetic (30 deg/sec) contractions (MVIC) on a Biodex dynamometer. MVIC testing was performed in a seated neutral upright posture. The upper thighs were stabilized using padded restraints to minimize extraneous movement. Each leg performed three consecutive knee extension and flexion (30°/second for concentric quadriceps and hamstring contractions) repetitions through the participants' full range of motion.

Participants completed run-and-cut trials at a 45-degree angle to both the right and left under four conditions: 1.) unanticipated with no cognitive task (NOCOG_UN), 2) anticipated with no cognitive task (NOCOG_AN), 3) unanticipated with a cognitive task (COG_UN), and 4) anticipated with a cognitive task (COG_AN). Cognitive load conditions were counterbalanced across participants, while anticipation trials were randomized.

In anticipation trials, the cut direction was known in advance. In unanticipated trials, the direction was cued by a visual arrow stimulus, triggered by a photoelectric timing gate 1.5 m before the force plate. In cognitive trials, participants performed a working-memory task while executing a 45° degree cut. A 5-digit sequence was presented (1 digit/sec) prior to beginning the run. Participants verbally recalled the sequence upon the researcher's requests and after completing the run-and-cut.

Participants first completed practice trials at a self-selected 'fast, game-like' pace. Practice continued until mean approach and exit speeds met the 3.25 m/s threshold and demonstrated consistency, defined as a coefficient of variation within ±2 SD of each participant's preferred speed. A trial was considered successful if the participant planted their full foot on the force plate, executed the correct cut direction, avoided cross-over steps, and achieved the minimum approach and exit speeds of 3.25 m/s. Trials were repeated if success criteria were not met or if EMG or motion-capture data were missing or corrupted due to equipment error. Each participant completed six successful trials per condition (three to each side).

To monitor fatigue and prevent overexertion, participants were asked their heart rate and self-reported rate of perceived exertion using a 1-10 scale (10 represents maximal exertion) every few trials.

Data Reduction & Analysis

Visual3D biomechanical software (v2025.01.2, C-Motion, Germantown, MD, USA) was used to process and analyze three-dimensional hip, knee, and ankle joint kinematic and kinetic variables. Raw marker coordinate and ground reaction force data were filtered using a fourth-order, zero lag, Butterworth low-pass filter with a cut-off frequency of 12 Hz [97,221]. Three-dimensional angular computations will be performed using Right-handed Cartesian segmental coordinate systems (x-y-z), where the x-axis represents the mediolateral direction, y represents anteroposterior, and z represents the longitudinal axis. Hip joint centers are defined at 25% of the distance from the ipsilateral to contralateral greater trochanter markers [222], knee joint centers are determined using the midpoint between femoral epicondyle markers [223], and ankle joint centers are determined using the midpoint between medial and lateral ipsilateral markers [224].

Raw EMG and MVIC data were processed using a fourth-order zero-lag order high pass Butterworth filter with a cutoff frequency of 10 Hz, followed by full wave rectification. The rectified signal was then low-pass filtered at 6 Hz to obtain the linear envelope[225]. MVIC EMG signals were smoothed with a 250-sample moving average filter (~125 ms at 2000 Hz) to reduce transient fluctuations and stabilize burst amplitude estimates. For each MVIC trial of consecutive isokinetic knee flexion/extension repetitions, the largest two amplitude bursts were identified, and the maximum amplitude within each burst was extracted. The average of these two maxima was taken as the peak MVIC value for normalizing the movement trials EMG data.

EMG trials were trimmed from 50 ms before to 100 ms after initial force plate contact. Mean normalized EMG amplitude was calculated within two windows: pre-contact (-50 ms to initial contact) and stance (0-100 ms post initial contact). Quadricep-to-hamstring (Q/H) co-activation ratio was also computed for both windows, defined as

$$Q/H_{ratio} = \frac{mean\ vastus\ medialis\ activity}{mean\ medial\ hamstring\ activity}.$$

A Q/H ratio greater than 1 indicates greater quadriceps relative to hamstrings activity, whereas values less than 1 indicate greater hamstrings relative to quadriceps activity.

Statistical Analysis

All data processing and analyses were completed in Python (v3.11.10) using custom scripts and the built in libraries: NumPy, pandas, SciPy, statsmodels, and pingouin. Linear mixed models (LMMs) were implemented to examine main and interaction effects of anticipation and cognitive load on each kinematic, kinetic and EMG variable. Each model followed a 2×2 factorial structure with *Anticipation* (Anticipated, Unanticipated) and *Cognitive Load* (Cognitive, No-Cognitive) as fixed factors, and a random intercept for *Participant* to account for within-subject dependence:

$$\begin{aligned} Y_{ijk} = {} & \beta_0 + \beta_1\left(Anticipation_j\right) + \beta_2(CognitiveLoad_k) \\ & + \beta_3\left(Anticipation_j \times CognitiveLoad_k\right) + u_{0i} + \varepsilon_{ijk} \end{aligned}$$

where:

- Y_{ijk} = dependent kinematic outcome for participant *i* under anticipation level *j* and cognitive load level *k*,
- β_0 = model intercept (grand mean),
- $\beta_1, \beta_2, \beta_3$ = fixed-effect coefficients for anticipation, cognitive load, and their interaction,
- u_{0i} = random intercept for participant *i*
- ε_{ijk}= residual error term, assumed ε_{ijk}~N(0,σ^2)

In the linear mixed model, the main effect reflects the difference between two conditions, combining across the other factor (anticipation and cognitive load). Models were estimated using maximum likelihood (ML) estimation. Fixed effect coefficients (β), standard errors, t-statistics, and p-values were extracted from model summaries. Post-hoc pairwise contrasts and estimated marginal means (EMMs) were computed using pingouin's, emmeans, and pairwise_tests functions, with Holm–Bonferroni correction for multiple comparisons. Model assumptions, including normality of residuals and homoscedasticity, were assessed using visual inspection of Q-Q plots and residual plots.

Results

Kinematic and Kinetic

At initial contact, a significant anticipation effect was found for frontal knee angle, with unanticipated trials having more knee abduction (mean difference = -13.79°, $p < .001$). Similarly, unanticipated trials demonstrated ~4° greater internal rotation at initial contact compared to anticipated trials ($\beta = 4.07$, $p < 0.001$). However, cognitive load had no significant main effects on sagittal, frontal, and transverse knee angles at the initial contact. Taken together, anticipation primarily influenced knee angles at initial contact, while cognitive load had no main effect. No significant interactions were observed for knee angles and moments.

In line with this, a significant anticipation effect was observed for peak knee flexion angle, with unanticipated trials averaging -60.2° compared to -58.2° in anticipated trials, reflecting a modest increase in knee flexion under unanticipated conditions ($\beta = –1.94$, $p = 0.04$). No significant effects were observed for cognitive load or the interaction in the sagittal plane. In addition, peak frontal and transverse knee angles showed no significant effects of anticipation, cognitive load effects, or their interaction. Overall, these findings suggest that anticipation influenced sagittal plane knee kinematics whereas frontal and transverse plane angles were unaffected.

In addition to knee angles, peak joint moments were evaluated. Unanticipated trials exhibited a significantly lower sagittal knee moment (2.72 vs. 3.39 Nm/kg), reflecting a reduction in extension moment under unanticipated conditions ($\beta = –0.67$, $p < 0.001$). In contrast, cognitive load had no significant effect on peak sagittal knee moment ($\beta = –0.06$, $p = 0.45$). A similar anticipation effect was observed in the frontal plane ($\beta = –0.10$, $p = 0.01$), with unanticipated trials showing lower peak moments than anticipated trials, while cognitive load had no effect ($\beta = 0.01$, $p = 0.89$). The cognitive load × anticipation interaction did not reach significance in either plane (sagittal: $\beta = 0.22$, $p = 0.06$; frontal: $\beta = 0.03$, $p = 0.57$). No significant main or interaction effects were found for transverse knee moments (all $p > 0.30$).

A significant anticipation effect was reported for both peak vertical ($\beta = -0.216$, $p = < 0.001$) and medial GRF ($\beta = 0.308$, $p = < 0.001$). Anticipation significantly affected mediolateral GRF, with anticipated trials averaging -0.41 BW compared to -0.72 BW in anticipated trials ($\beta =$

0.308, p = < 0.001). Interestingly, vertical GRF was significantly lower in unanticipated trials with a slight decrease (β = -0.216, $p < 0.001$), reflecting peak magnitude of 2.06 BW compared to 2.28 BW for anticipated conditions. A significant interaction between cognitive load and anticipation was found for anteroposterior GRF (β = −0.12, $p = .006$). Post hoc comparisons confirmed differences between anticipation conditions in both the no-cognitive load (β = –0.308, $p < 0.001$) and cognitive load conditions (β = –0.188, $p < 0.001$). Pairwise contrasts further indicated differences between NOCOG_UN and COG_AN (β = 0.272, $p < 0.001$), NOCOG_AN and COG_UN (β = -0.224, $p < 0.001$), and between NOCOG_UN and COG_UN (β = 0.085, p = 0.006). Overall, anticipation consistently influenced mediolateral GRF, while cognitive load effects were limited to interaction patterns. No significant anticipation or cognitive effects or interactions were found for peak anteroposterior GRF.

EMG Activation

During the pre-initial contact phase, anticipation had a significant effect on hamstring activation (β = -0.09, $p = 0.032$), with unanticipated trials averaging 0.89 compared to 0.79 in anticipated trials. No significant anticipation, cognitive, or interaction effects were found for mean medial hamstring activation during stance for vastus lateralis activation at either phase. In contrast, cognitive load significantly affects the Q/H ratio. At pre-initial contact, cognitive conditions showed higher ratios than no cognitive conditions (0.96 vs 0.73, β = 0.23, $p = 0.028$), reflecting a shift toward greater quadriceps activation. Conversely, at stance phase, cognitive load reduced the ratio (2.60 vs 3.67; β = -1.07, $p = 0.027$), reflecting a relative increase in hamstring activation. Overall, anticipation influenced hamstring activation, while cognitive load altered quadriceps-to-hamstring coordination across phases, with no other significant EMG effects observed.

Discussion

This study aimed to examine how additional cognitive load, applied through a working-memory digit recall task, influenced lower extremity mechanics during anticipated and unanticipated cutting. Contrary to expectations, no significant anticipation x cognitive load interactions were observed for knee joint angles or moments. However, a significant interaction emerged for peak anteroposterior GRF, with anticipated trials producing greater posterior GRF magnitudes. This pattern was consistent under both cognitive and non-cognitive conditions, though the difference between anticipated and unanticipated was greater under no-cognitive load conditions.

Anticipation effects were also observed for vertical GRF and peak internal knee extensor moments. Together, these results suggest that anticipated trials elicited a more quadriceps-driven strategy when the movement was known in advance. Similar braking-dominant patterns have been reported in anticipated landings, where increased quadriceps activation and reduced knee flexion contribute to higher impact forces and loading rates (Decker et al., 2003; Schmitz et al., 2007). Consistent with this, anticipation was also associated with greater peak knee adduction moment, suggesting increased frontal plane loading. This finding aligns with prior studies, supporting the notion that anticipation influences multiplanar knee loading patterns relevant to ACL injury risk [227-229]. Overall, these results suggest that when the movement was known in advance, participants demonstrated a proactive, anticipatory control strategy, characterized by the greater breaking magnitudes and reduced knee flexion[230,231].

In contrast, unanticipated conditions showed greater knee flexion angles, along with increased internal rotation angle at initial contact. Taken together with the proactive braking strategy observed in anticipated trials, these findings indicate two distinct adaptations with different ACL-relevant implications. The stiffer knee position adopted in anticipated trials coincided with higher adduction moments observed, a notable pattern previously identified as a key ACL injury mechanism [8,22,81]. However, recent work by Ebner, Granacher and Gehring [232] has systematically reviewed the implications of that in the change-of-direction tasks (e.g., cutting), noting several studies reported greater knee flexion angles in unanticipated conditions [233-235]and peak internal knee adduction and extension moments and in anticipated conditions [228] [227,229]. Moreover, interpretation of these data warrant caution, as unanticipated cutting should not be assumed to equate to safer mechanics. ACL injury is a multi-faceted phenomenon involving neuromuscular control, joint mechanics, and cognitive influences. In this context, our EMG results help to expand this perspective.

It is important to note that EMG measures were averaged across the phase and cannot be directly tied to discrete joint events. As such, EMG and joint mechanic outcomes should be considered as independent adaptations. Still, the phase-specific Q/H ratio findings suggest that cognitive load altered muscle activation, with evidence of relative reduced quadriceps activity during stance. This shift points to a potential cognitive-motor trade-off, where working-memory demands influence neuromuscular balance. Such effects highlight the importance of investigating phase-specific activation with time-series approaches that align muscle activation and kinematics temporally.

In line with this, it is important to note that group-level analyses implemented in this study can mask unique individual movement patterns, which are of equal importance when evaluating ACL injury risk. Participants were instructed to cut at a self-selected, game-like speed to replicate competition demands as best as possible in a controlled laboratory setting, and approach velocity was subsequently included as a covariate in the statistical models. While this approach controlled for variability, it is likely that participants may not have performed at their truest-highest intensities. Prior work has suggested that 45° cuts typically elicit ACL-relevant mechanics and knee joint moments only when approach speeds reach at least ~4 m/s [236].

Beyond approach speed, several other methodological factors warrant caution in interpreting these findings. Cuts were performed under controlled laboratory conditions with standardized instructions and an added working-memory task, which likely reduced the reactive demands typical of faster, unpredictable game scenarios. The cognitive task itself may not have been sufficiently difficult to draw resources away from motor control, and the cutting task may not have imposed enough motor challenge to elicit strong competition for attentional resources. These design choices may help explain the absence of robust interaction effects. In addition, reaction time was not measured yet is likely to differ between anticipated and unanticipated trials, influencing the observed movement patterns. Finally, while joint angles, moments, and Q/H ratios provided insight into mechanics and muscle balance, they may not capture more subtle neuromuscular adaptations, such as timing, variability, or coordination, which could be critical in understanding dual-task injury risk. Interaction effects may therefore exist in other dimensions of movement behavior that were beyond the scope of this analysis but remain important for characterizing individual responses to cognitive–motor challenges.

Taken together, these considerations underscore that while the present paradigm effectively isolated cognitive–motor interactions under controlled conditions, it does not fully capture the dynamics of sport-like performance. Future work should implement more ecologically valid task demands (e.g., higher speeds, more reactive environments), increase cognitive task complexity, and integrate continuous measures such as reaction time or time-series EMG to capture a fuller picture of how anticipation and cognitive load shape movement strategies.

These findings diverge from prior work reporting increased quadriceps loading under unanticipated conditions, likely reflecting differences in experimental context. In this dual-task paradigm, participants performed cuts in a controlled laboratory setting with standardized instructions and an added working-memory task. While these features allowed for isolating cognitive-motor interactions, they may have reduced the reactive demands typical of faster, game-like scenarios. In addition, participants performed cuts at a self-selected running speed that was monitored within an acceptable range (>3.25 m/s), ensuring consistent approach velocity across trials. However, reaction time was not measured, and it is likely that participants responded more slowly in unanticipated conditions, consequently influencing the observed movement patterns.

Taken together, these factors highlight that while the present design isolated cognitive-motor interactions under controlled conditions, future work should incorporate measures such as reaction time, and more ecologically valid task demands to full capture how anticipation and cognitive influence movement strategies.

Conclusion

This study showed anticipated trials produced greater posterior braking forces and larger knee extensor moments, whereas unanticipated trials involved greater knee flexion angles, both reflecting distinct strategies for managing load. While prior studies have reported reduced knee flexion and elevated ACL injury risk during unanticipated tasks, the present findings diverge from that pattern. These differences may reflect the dual-task paradigm employed here to introduce additional cognitive demand, as well as characteristics of the study population, encompassing physically active but not elite female athletes. Such factors may limit direct comparison to competitive contexts and warrant cautious interpretation. It should also be recognized the laboratory-based paradigms cannot fully replicate the complexity of game-like environments. Both competitive play and practice involve varying levels of stress fatigue, and physical interactions with other opponents – all of which are difficult to reproduce and measure, though remain relevant to ACL injury occurrence.

Future work should progress this research domain by incorporating sport-specific, beyond decision making alone, to include measures such as reaction time and more complex cognitive loads. Additionally, expanded taxonomies of dual task demands may provide a more comprehensive framework for examining how cognitive and motor processes interact. Such approaches will better capture the cognitive-motor challenges of game-like settings, enhancing ecological validity and refining return-to-play protocols that more effectively address both the motor and cognitive components of ACL injury risk.

Table 1. Group mean (± SD) knee joint angles (°) at initial contact and global peak.

		NOCOG_AN	NOCOG_UN	COG_AN	COG_UN
Initial Contact	Flexion/Extension[a]	-28.62 ± 9.65	-42.90 ± 15.12	-27.68 ± 8.76	-40.43 ± 16.01
	Abduction/Adduction	-1.01 ± 3.75	-1.69 ± 3.77	-1.01 ± 3.75	-2.11 ± 3.56
	External/Internal Rotation[a]	-1.85 ± 7.95	2.58 ± 6.15	-1.72 ± 6.19	3.32 ± 8.20
Peak	Flexion/Extension[b]	-58.57 ± 7.06	-60.33 ± 8.35	-57.32 ± 5.82	-60.13 ± 9.90
	Abduction/Adduction	-8.06 ± 5.03	-8.17 ± 4.86	-7.72 ± 3.90	-7.99 ± 3.67
	External/Internal Rotation	13.11 ± 4.81	13.41 ± 4.70	13.41 ± 3.21	13.17 ± 4.83

Note. Values represent the corresponding (-/+) joint angle convention listen in each row.

[a]Significant main effect of anticipation ($p < 0.001$); [b]Significant main effect of anticipation ($p < 0.05$)

Table 2. Group mean (± SD) peak knee joint moments (Nm/kg).

	NOCOG_AN	NOCOG_UN	COG_AN	COG_UN
Extension[a]	3.42 ± 0.53	2.72 ± 0.62	3.31 ± 0.61	2.87 ±0.64
Adduction[b]	0.50 ±0.30	0.37 ±0.36	0.48 ± 0.31	0.41 ± 0.31
External Rotation	-0.18 ± 0.13	-0.16 ± 0.11	-0.17 ± 0.15	-0.24 ± 0.67

Values represent the internal joint moment. [a]Significant main effect of anticipation ($p < 0.001$); [b]Significant main effect of anticipation ($p < 0.05$)

Table 3. Group mean (± SD) peak GRF (BW).

	NOCOG_AN	NOCOG_UN	COG_AN	COG_UN
Vertical GRF[a]	2.28 ± 0.26	2.06 ± 0.27	2.27 ± 0.27	2.14 ± 0.30
Anteroposterior GRF[c]	-0.74 ± 0.20	-0.40 ± 0.27	-0.68 ± 0.21	-0.49 ± 0.36
Mediolateral GRF	0.56 ± 0.77	0.60 ± 0.72	0.59 ± 0.76	0.56 ± 0.72

[a]Significant main effect of anticipation ($p < 0.001$); [c]Significant anticipation × cognitive load interaction ($p < 0.05$).

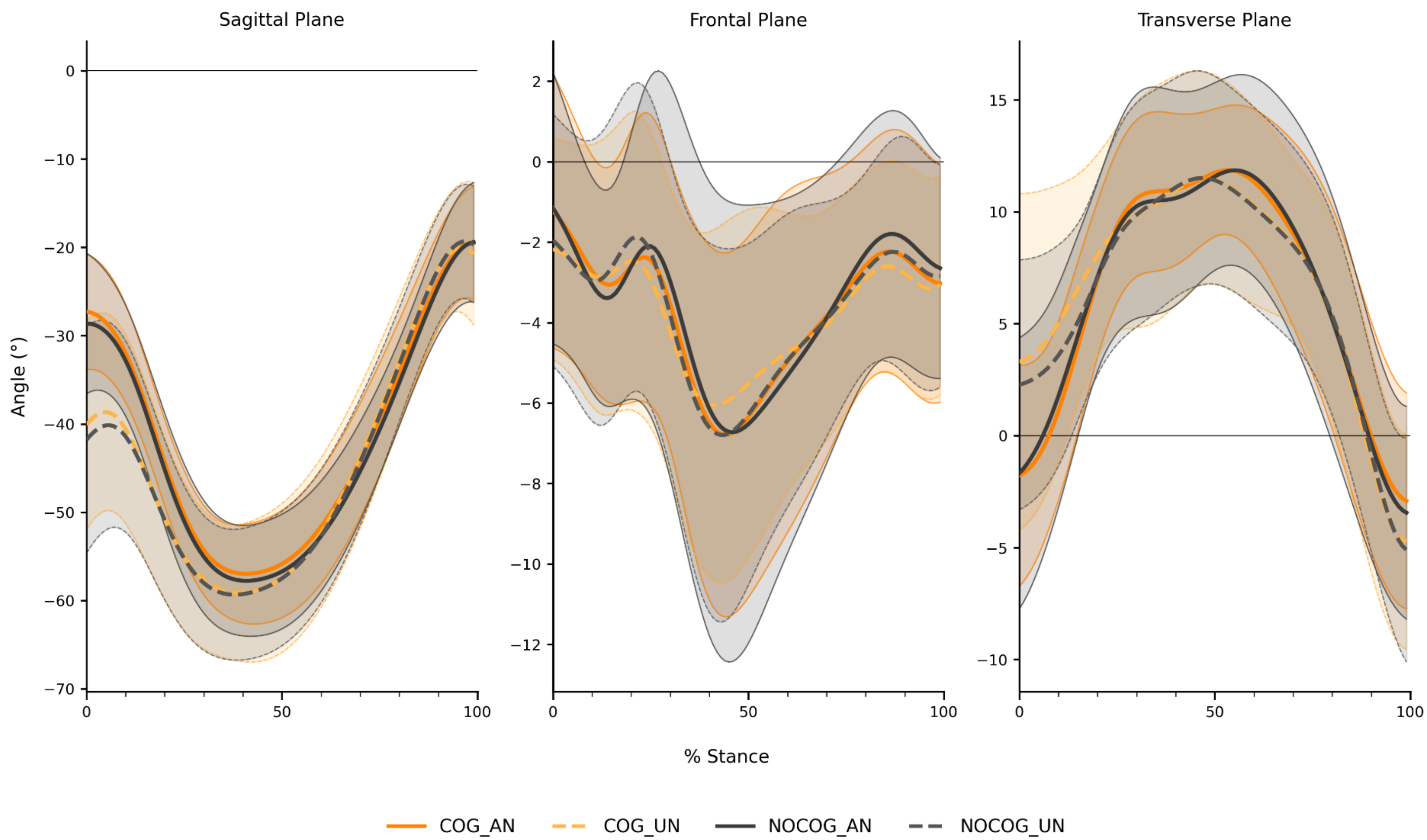

Figure 1. Group mean ensemble curves of knee joint angles across 0–100% stance, with shaded regions representing ±1 standard deviation.

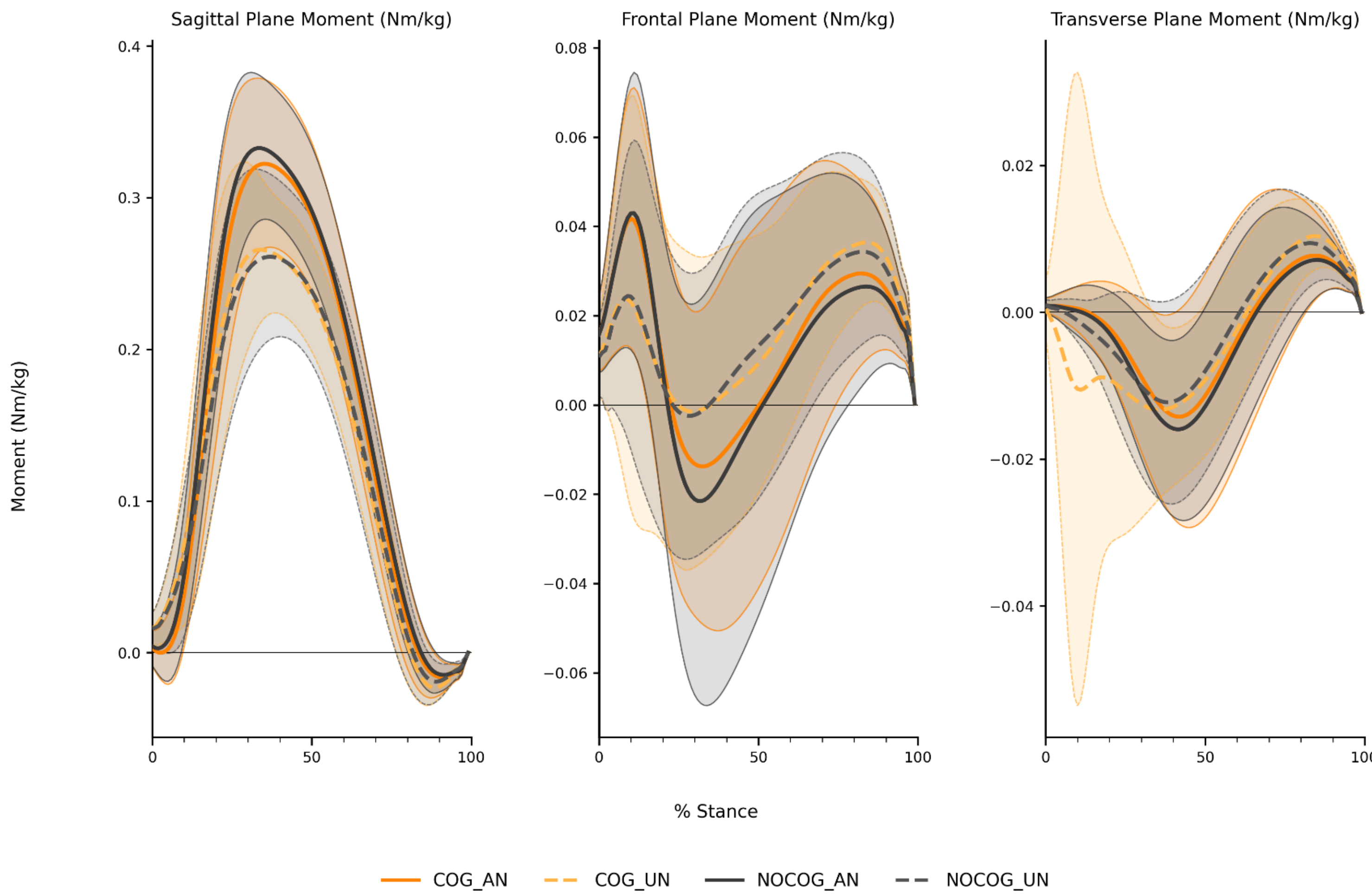

Figure 2. Group mean ensemble curves of knee joint moments across 0–100% stance, with shaded regions representing ±1 standard deviation.

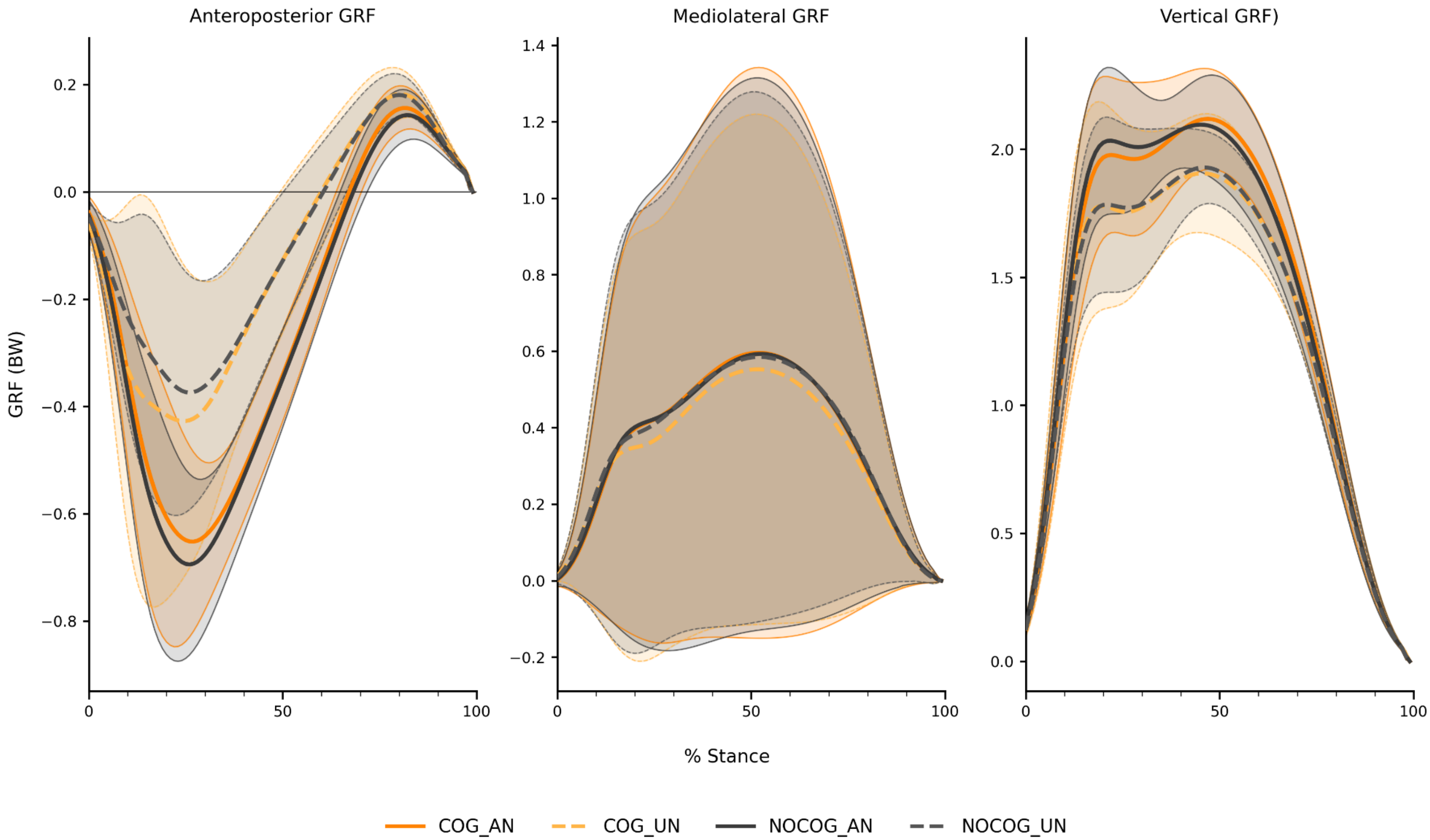

Figure 3. Group mean ensemble curves of ground reaction force across 0–100% stance, with shaded regions representing ±1 standard deviation.

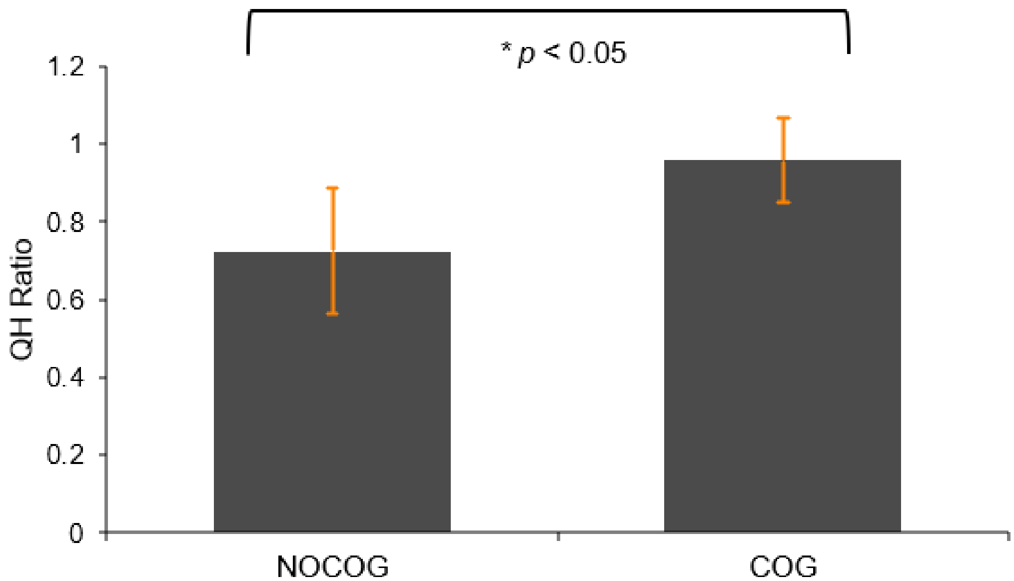

Figure 4. Estimated marginal means (±SE) of the quadriceps–hamstring (QH) coactivation ratio averaged across the 50 ms before force plate contact under(NOCOG) and dual-task (COG) conditions, averaged across anticipated and unanticipated landings. A significant main effect of cognitive load was observed (p=0.027).

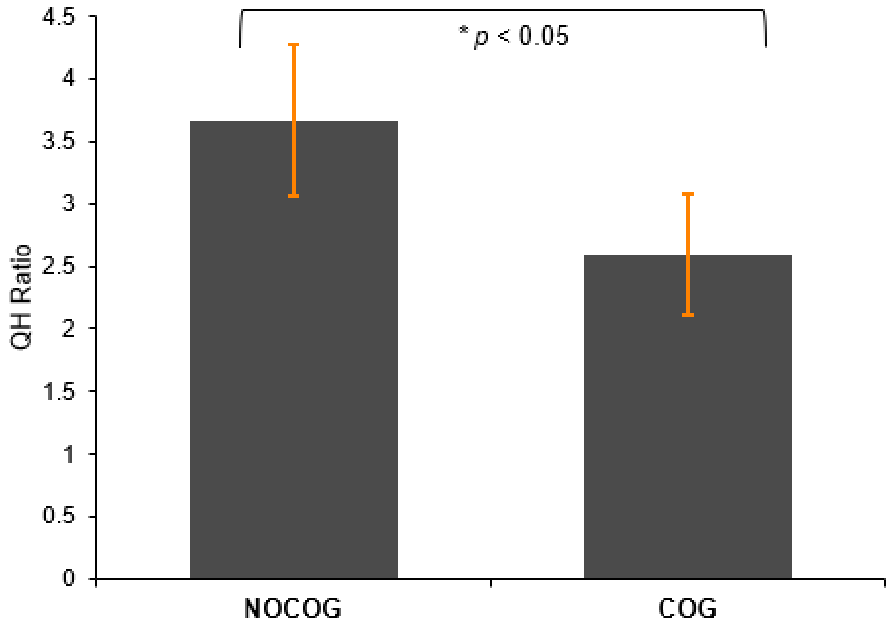

Figure 5. Estimated marginal means (±SE) of the quadriceps–hamstring (QH) coactivation ratio averaged across the window from initial force plate contact +100 ms under (NOCOG) and dual-task (COG) conditions, averaged across anticipated and unanticipated landings. A significant main effect of cognitive load was observed (p=0.028).

Chapter 5: Cognitive Motor Interference – Dual Task Cost Analysis

Abstract

This study examined the effects of cognitive-motor interference (CMI) during run-and-cut tasks. Dual-task paradigms were used to evaluate how increased cognitive load, induced through a working memory digit recall task influences cognitive and motor performance. Participants performed a serial digit-recall task under Baseline (cognitive only), Anticipated, and Unanticipated cutting conditions. Sequence accuracy was analyzed using a one-way repeated measures ANOVA. Dual-task effect (DTC) values were calculated to represent relative performance change from Baseline to dual-task conditions, with a paired-samples t-test comparing DTC between Anticipated and Unanticipated conditions. Secondary analyses examined associations between digit recall performance and kinematics and kinetics using Pearson's correlations.

No significant differences in sequence, accuracy or DTC between conditions were observed. In the anticipated condition, DTC was positively associated with peak knee external rotation and adduction moments, as well as anteroposterior ground reaction force. In the unanticipated condition, DTC showed the strongest relationship with peak knee external rotation moment and a moderate association with peak knee flexion angle.

Overall, the findings highlight that even when group-level motor performance is maintained, cognitive efficiency may be compromised under dual-task conditions, offering insight into attentional mechanisms contributing to movement variability and consequent injury risk.

Introduction

Cognitive processing plays an important role in athletic performance and is increasingly recognized as a factor influencing injury risk. This is particularly relevant in sports that demand rapid motor control, where anterior cruciate ligament (ACL) injuries occur during cutting or landing while athletes simultaneously regulate attention and movement execution. Extensive research on ACL injury risk has been centered around mechanical and neuromuscular contributors. Specifically, limited knee flexion angle, increased knee abduction angle, and increased internal knee adduction moment have been shown to increase anterior tibial shear forces and strain on the ligament, heightening injury susceptibility [8,23,54,72,229]. While these factors describe mechanical conditions that predispose the ACL to failure, they do not capture the multifaceted nature of injury. Often, athletes must allocate attention across several competing demands, involving play recall, opponent tracking, and decision making, all while simultaneously maintaining proper motor coordination. This interplay between cognitive demands and motor control operates within a broader framework of ACL injury risk that extends beyond traditional mechanical explanations.

Within this scope, dual-task paradigms involve simultaneous engagement in motor and cognitive processes. Anticipated and unanticipated movement paradigms, which require athletes to respond to directional or visual cues, often represent implicit dual-task conditions in ACL research. However, what is known about ACL injury risk under these conditions primarily reflects decision-making or cue recognition as the sole source of cognitive demand. This narrow focus limits the ecological validity, as sport performance involves imposed additional cognitive load such as working memory maintenance, tracking multiple stimuli, and updating decisions under time pressure. Previous work examining cutting, landing, and gait tasks under cognitive load has demonstrated altered lower-limb mechanics, including reduced knee flexion angles and increased frontal-plane moments [237] [238,239] [240]. In ACL-reconstructed populations, added cognitive load has been shown to amplify limb asymmetries and delay muscle activation [239]. Despite these findings, the level of cognitive load effects has varied across studies, likely due to differences in task complexity, cue timing, and the type of cognitive demand imposed. A systematic review by Ebner, Granacher and Gehring [232] highlighted inconsistencies, demonstrating that anticipation and dual-task manipulations do not uniformly alter lower-limb mechanics. Such variability suggests that attentional effects on movement may depend on both methodological design and individual strategy.

The interaction between concurrent cognitive processing and motor execution is referred to as cognitive-motor interference (CMI). Attentional demands describe the cognitive resources required to manage and allocate resources. Correspondingly, CMI reflects the degree to which those shared resources compete during simultaneous cognitive and motor processing. In cognitive-motor paradigms, performance changes may occur in either domain depending on how attentional resources are prioritized. Although this framework of attention has been explored in motor learning and gait research [201,241,242], CMI has not been explicitly classified in ACL injury-related studies. Most investigations infer interference indirectly through changes in movement mechanics, without concurrently assessing cognitive performance or the trade-off between domains. To overcome this limitation, dual-task cost or effect (DTC) provides a quantitative method to characterize the degree of CMI by comparing performance between single- and dual-task conditions.

Dual task cost represents the relative change in the separate cognitive and motor outcomes, with positive and negative values corresponding with performance improvement or decrement. When combined with the dual-task taxonomy proposed by Plummer and Eskes [39], outcomes can be classified as facilitation, interference, or stability in performance. Facilitation reflects improved performance under dual-task conditions, interference indicates performance decline, and stability denotes minimal or no change from single-task performance. This is particularly relevant to ACL injury as cognitive load may alter neuromuscular coordination and joint mechanics associated with injury risk [243-245]. Quantifying and categorizing distinct patterns of interference provides insight into how varying task demands elicit different cognitive-motor attentional strategies.

Empirical evidence from previous postural and gait studies illustrate how interference patterns manifest in practice [32,246]. When motor stability is challenged, individuals tend to prioritize motor control, yielding reduced cognitive performance [241,242,247]. Conversely, when attention is directed toward cognitive performance, such as decision making or information recall, performance decrements have been observed at the expense of the motor domain [248]. Together, these findings demonstrate that attentional resource allocation is dynamic and context specific, emphasizing the need to examine such interactions in complex athletic movements. However, to our knowledge, the quantification and classification of CMI have not yet been applied within the context of ACL injury. This is particularly relevant, as divided attention may alter movement control and strategies associated with injury risk.

Therefore, the purpose of this study was to quantify cognitive-motor interference during run-and-cut tasks using a cognitive-motor dual-task paradigm. It was hypothesized that imposed working memory load would modify knee joint mechanics associated with ACL injury risk, specifically by reducing knee flexion angles and increasing internal knee adduction moments. Furthermore, classifying the resulting patterns of CMI provides a framework to interpret how cognitive load influences motor performance in individual movement strategies.

Methods

Participants

Eighteen participants were recruited; however, data from one participant were excluded due to data collection error, resulting in a final sample of seventeen healthy females (22.35 ± 2.78 years; height: 1.69 ± 0.05 m; body mass: 62.96 ± 7.54 kg) reported no history of lower extremity injury within the past six months and no prior lower extremity surgery. Inclusion criteria required that participants engaged in regular physical activity, defined as meeting American College of sports Medicine guidelines of at least 150 minutes of moderate-intensity or 75 minutes of vigorous intensity activity per week. Prior to data collection, experimental procedures received ethical approval from the university's Institutional Review Board and written informed consent was obtained.

Instrumentation

A Trigno^{TM} Wireless EMG system (2000 Hz, Delsys, Inc., Natick, MA, USA) measured muscle activity during the cutting tasks. MVIC testing was conducted using a dynamometer (Biodex Medical Systems, System 4 Pro, NY, USA). Raw marker kinematic data will be collected using a 12-camera motion capture system (200 Hz, Vicon, Centennial, CO, USA). Twenty-seven retro-reflective markers were placed on the manubrium below the sternal notch, C7, L1, and bilaterally on the right and left acromion processes, iliac crests, anterior superior iliac spines (ASIS), posterior superior iliac spines (PSIS), greater trochanters medial and lateral femoral epicondyles, medial and lateral malleoli, tips of the second toes, and the first and fifth metatarsophalangeal joints. Clusters of four non-collinear markers on rigid thermoplastic shells were secured using neoprene straps to the posterior trunk, posterior pelvis, and bilaterally to the lateral thighs and shanks, and to the posterior surface of both shoe heels. Once a standing static calibration trial is recorded, the 27 anatomical markers were removed. Synchronous ground reaction force data were collected via AMTI force plates (2000 Hz, AMTI, Inc., Watertown, MA, USA). LabVIEW VI was used to present the visual stimulus on a computer monitor placed 3 m from the force plate. The Polar Verity Sense optical heart rate sensor measured heart rate (Polar Electro, USA). Statistical analysis will be completed in SPSS (IBM SPSS Statistics, Version 29, Armonk, NY, USA) and Python (3.10, Python Software Foundation, 2024).

Experimental Procedures

First, participants completed a baseline digit recall test while sitting in front of a television screen. The baseline cognitive test consisted of twelve unique sequences of five consecutive digits on the screen – the same screen that was later used for the movement trials. Each sequence was presented one digit at a time with each digit displayed on a separate slide of a PowerPoint presentation. Blank screens were used to separate the sequences. Participants were instructed to recall the sequences verbally, only when prompted by the researcher. Pauses of 10 to 30 seconds were provided between sequences to replicate the memory retention period that would occur during movement trials. Responses were recorded by the researcher on a standardized data collection sheet.

Next, electromyography (EMG) sensors were placed on the right and left vastus medialis and medial hamstring in accordance with the SENIAM guidelines [220]. Participants then completed a 3-minute treadmill warm-up at a self-selected pace, followed by maximal voluntary isokinetic contractions (MVIC) on a Biodex dynamometer. MVIC testing was performed in a seated neutral upright posture. The upper thighs were stabilized using padded restraints to minimize extraneous movement. Each leg performed three consecutive knee extension and flexion (30°/second for concentric quadriceps and hamstring contractions) repetitions through the participants' full range of motion.

Participants completed run-and-cut trials at a 45-degree angle to both the right and left under four conditions: 1.) unanticipated with no cognitive task (NOCOG_UN), 2) anticipated with no cognitive task (NOCOG_AN), 3) unanticipated with a cognitive task (COG_UN), and 4) anticipated with a cognitive task (COG_AN). Cognitive load conditions were counterbalanced across participants, while anticipation trials were randomized.

In anticipation trials, the cut direction was known in advance. In unanticipated trials, the direction was cued by a visual arrow stimulus, triggered by a photoelectric timing gate 1.5 m before the force plate. In cognitive trials, participants performed a working-memory task while executing a 45° degree cut. A 5-digit sequence was presented (1 digit/sec) prior to beginning the run. Participants verbally recalled the sequence upon the researcher's requests and after completing the run-and-cut.

Practice trials of the run-and-cut were completed at a self-selected 'fast, game-like' pace. Practice continued until mean approach and exit speeds met the 3.25 m/s threshold and demonstrated consistency, defined as a coefficient of variation within ±2 SD of each participant's preferred speed. A trial was considered successful if the participant planted their full foot on the force plate, executed the correct cut direction, avoided cross-over steps, and achieved the minimum approach and exit speeds of 3.25 m/s. Trials were repeated if success criteria were not met or if EMG or motion-capture data were missing or corrupted due to equipment error. Each participant completed six successful trials per condition (three to each side).

To monitor fatigue and prevent overexertion, participants were asked their heart rate and rate of perceive exertion using a 6–20-point Borg scale (20 represents maximal exertion) every 4 trials. Between every trial sufficient rest time was provided, determined by the participant.

Data Reduction & Analysis

Sequence Accuracy. As digit sequence length was identical across conditions, recall performance was quantified as the percentage of correctly recalled digits per trial, with credit given only when digits were recalled in the correct order. Sequence order was accounted for to capture the overall working memory load rather than serial recall.

$$Sequence\ accuracy\ (\%) = \frac{number\ of\ correct\ digits\ recalled}{sequence\ length} \times 100.$$

For example, if the sequence presented was *12345* and the subject recalled *12435*, three digits were recalled in the correct position, resulting in a sequence accuracy of 60%. For each participant, trial-level accuracies were averaged for the single-task baseline test and within each cognitive condition (COG_AN, COG_UN) to produce a single accuracy score per condition. Participant trial-level accuracies served as the dependent variable in the repeated-measures ANOVA.

Dual Task Cost. To quantify the relative change in performance from single to dual task conditions, dual task costs (i.e., effects) (DTC) were calculated using percentage accuracy values from both dual task and single task (baseline) conditions for each participant.

$$DTC\ (\%) = \frac{Dual\ task\ accuracy - Single\ task\ accuracy}{Single\ task\ accuracy} \times 100.$$

Cognitive Motor Interference. Further evaluation of cognitive motor interference (CMI) was classified following the framework outlined by Plummer [32]. Cognitive and motor performance metrics were compared between single and dual task conditions.

Data processing and classification were completed in Python using custom analysis scripts. For each biomechanical metric trial-level data were averaged within condition (NOCOG_AN as the baseline, COG_AN, and COG_UN). To preserve the measured metrics' polarity and physical meaning, changes in motor performance were quantified as $\Delta_{motor} = \bar{X}_{dual\ task\ conditoin} - \bar{X}_{single\ motor\ task\ condition}$. To determine whether a change reflected a meaningful decrement or improvement, thresholds were established for each metric and defined as two standard deviations (±2 SD) of the baseline single task anticipated condition values, calculated separately for each outcome variable. A threshold of ±2 SD was selected to reduce the likelihood of misclassifying normal within-subject variability as meaningful motor change [249]. Motor changes within this range were classified as "no change," decreases beyond –2 SD as "motor interference," and increases beyond +2 SD as "motor facilitation." These values were then examined with respect to the cognitive DTC scores to classify the interference patterns established as: *no interference, motor facilitation, cognitive facilitation, mutual facilitation, motor-related cognitive interference, cognitive-related motor interference, motor-priority trade-off, cognitive-priority trade-off, and mutual interference.*

Statistical Analysis

All analyses were performed in Python (v3.11.10) using the Pingouin (pg) statistics package [250]. Digit recall accuracy was analyzed using a one-way repeated-measures analysis of variance (ANOVA), with condition (NOCOG_AN, COG_AN, and COG_UN) as the within-subject factor. Assumptions of sphericity was evaluated using Mauchly's test with Greenhouse-Geisser corrections applied. Effect sizes are reported as partial eta squared (ηp^2). Where appropriate, pairwise comparisons were conducted using paired simple t-tests with Bonferroni correction to adjust for multiple comparisons. In addition, a paired-samples t-test compared DTC between anticipated and unanticipated conditions. Effect sizes were reported using Cohen's *d*.

As a secondary analysis, associations between digit recall performance and biomechanical outcomes (initial contact knee angles, peak ground reaction force, and peak knee angles) were examined using Pearson's correlations. To account for individual baseline differences in recall accuracy, DTC values were used in the analysis. Correlations are reported as r values with significance set at $p < 0.05$. Effect sizes were categorized as small ($r = .10$), medium ($r = .30$), and large ($r \geq .50$) following Cohen's conventions.

Results

Digit Recall Accuracy

Participant digit recall accuracy scores are shown in **Table 42**, **Appendix G**. A one-way repeated-measures ANOVA revealed no significant main effect of condition on digit recall accuracy was revealed ($F(2, 32) = 0.12$, $p = .885$, $\eta p^2 = .003$). Mauchly's test indicated that the assumption of sphericity was violated ($\chi^2(2) = 7.41$, $p = .023$); however, applying the Greenhouse–Geisser correction did not alter the result (corrected $p = .817$).

Cognitive Dual Task Cost

A paired-samples *t*-test showed no evidence of a difference in DTC between anticipated and unanticipated conditions ($t(16) = 0.22$, $p = 0.826$). The confidence interval (–3.48 to 4.30) suggests that the true difference was likely negligible. Consistent with this, the effect size was small (Cohen's $d = 0.04$).

Biomechanical Outcomes

As detailed in Chapter 4, linear mixed model (LMM) analyses examined all four experimental conditions, (COG_AN, COG_UN, NOCOG_AN, NOCOG_UN) and revealed no significant main effects of cognitive load on knee joint mechanics (see **Tables 1-3, Appendix. Chapter 4**). To provide descriptive context for the dual-task analyses, the present chapter focuses on three primary conditions: NOCOG_AN (baseline) and the cognitive anticipated and unanticipated conditions (COG_AN, COG_UN). Descriptive group means across these conditions are summarized in **Tables 4-6, Appendix. Chapter 5**. Given the absence of cognitive effects at the group level, we conducted exploratory analyses to evaluate potential dual-task influences. These included (1) bivariate correlations between cognitive DTC and participant-level mean biomechanics across all 12 variables (initial contact knee angle, peak knee angles, peak knee moments, and peak GRFs), analyzed separately for anticipated and unanticipated cognitive conditions, separately; and (2) classification of participants into CMI patterns based on paired motor and cognitive DTC values.

Dual-task Cost

Bivariate Pearson's correlation analyses were conducted to examine the associations between biomechanical measures and DTC, with more positive DTC values indicate less cognitive performance interference (i.e., better preservation or facilitation of recall accuracy). In the anticipated condition, DTC was significantly associated with greater peak knee external rotation moment ($r = 0.61$, $p < 0.01$), peak knee adduction moment ($r = 0.60$, $p < 0.01$), and anteroposterior GRF ($r = 0.53$, $p < 0.05$). In the unanticipated condition, DTC was most strongly associated with peak knee external rotation moment ($r = 0.68$, $p < 0.01$) and was moderately correlated with peak knee flexion angle ($r = 0.50$, $p = 0.042$). No other associations reached significance. Scatterplots depicting the correlations between knee joint variables and DTC are shown in **Figures 60** for COG_AN and **Figure 61** for COG_UN (**Appendix G**).

Cognitive Motor Interference Classification.

Across all participants and metrics, CMI classifications were assessed in both anticipated and unanticipated cutting conditions. **Figure 8 (Appendix. Chapter 5)** presents the distribution of classifications across all metrics, separated by anticipated and unanticipated conditions. Percentages reflect the proportion of total classifications assigned to each CMI category, pooled across all participants and all evaluated metrics. Across metrics, the most frequent classifications corresponding with the COG_AN condition were *cognitive-facilitation* (40.7%), followed by *no dual task interference* (29.4%) and *motor-related cognitive interference* (29.4%). Interestingly, for the COG_UN condition, no dual-task interference was the most common outcome (36.8%), followed by motor-related cognitive facilitation (30.9%), and cognitive facilitation (20.6%).

To examine ACL-relevant mechanics, specifically, classifications were summarized for peak knee flexion angle, peak abduction angle, and peak knee adduction moment (**Table 7, Appendix. Chapter 5**). Across anticipated trials, all three variables were most often classified as cognitive facilitation (41%), followed by motor-related cognitive interference and no interference (29% each). In unanticipated trials, the distribution of classifications shifted toward no dual-task interference for peak knee flexion angle and knee adduction moment (35% each) and knee abduction angle (41.2%), while cognitive facilitation was less frequent (23-24%). Participant classification patterns displayed greater variety, with smaller proportions of motor-priority trade-offs and motor facilitation (<6%).

Notably, classifications for the metrics that demonstrated significant correlations with cognitive DTC (e.g., external rotation moment, mediolateral GRF) paralleled these patterns, with facilitation most common in anticipated trials and no interference more frequent in unanticipated trials. The corresponding summary tables for all biomechanical variables are provided in **Tables 43–54** (**Appendix H)**, detailing the number and percentage of participants classified within each cognitive–motor interference (CMI) category for both anticipated and unanticipated conditions.

Discussion

Understanding how cognitive demands influence movement strategies is critical for advancing knowledge of ACL injury and risk. This study tested the hypothesis that dual task performance with a cognitive component of increased working memory demands would alter lower extremity mechanics during run-and-cut maneuvers in ways associated with increased ACL loading. Additionally, interference classifications were expected to provide insight into how athletes prioritize motor and cognitive performance under dual task demands.

The results did not support the hypothesis that cognitive load would alter biomechanics, as no significant main effects were observed (**Appendix F**). Likewise, no significant differences in digit recall accuracy were observed across baseline (NOCOG_AN, COG_AN, and COG_UN conditions. Cognitive DTC also did not differ meaningfully between anticipation conditions. The absence of significant changes under cognitive load aligns with previous findings reporting minimal or inconsistent effects during dual task cutting or landing [203,232]. Conversely, other studies have demonstrated increased knee adduction moments and reduced flexion angles under unanticipated load [26,227,234]. While these findings suggest minimal group-level effects, they do not fully capture the trade-offs that emerge when participants navigate concurrent cognitive-motor tasks. To gain further insight into how the interplay of concurrent cognitive and motor demands show up, exploratory analyses assessed correlations between cognitive cost with the kinematic and kinetic variables.

Correlations provided a continuous view of how cognitive load was associated with lower extremity mechanics in this paradigm, while CMI classifications offered a complementary categorical lens. Together these approaches revealed that participants adopt distinct dual-task strategies, ranging from facilitation to interference, notably differing between anticipated unanticipated conditions. Participants that maintained or improved cognitive performance under the anticipated condition tended to show relative greater knee adduction moments and reduced external rotation moments, as well as reduced braking GRF, compared to their counterparts who demonstrated poorer preservation of cognitive performance. In the unanticipated condition, participants who preserved cognitive performance showed a consistent pattern of smaller knee flexion angles and external rotation moments near neutral than those with lower DTC scores. Notably, greater adduction moments have been identified as a joint factor associated with elevated ACL injury risk [22,81]. However, these findings should be interpreted with caution. Rather than indicating a uniform increase in risk, the observed relationships suggest a potential trade-off. Preserving cognitive performance may coincide with relatively greater frontal plane loading but also reduction in transverse plane moments and braking forces, reflecting the multidimensional nature of motor adaptations under cognitive load.

Given that ACL injuries often occur in unpredictable environments with concurrent cognitive and motor demands, it's critical to understand how athletes differ in the strategies elicited to manage dual task demands during cutting. For knee flexion angles, abduction angles, and adduction moments within the anticipated condition, participants most often improved cognitive performance with no motor change or showed declines in cognitive accuracy while preserving motor performance, with a similar proportion showing no interference. In contrast, the unanticipated condition more frequently reflected stable performance in both domains (i.e., no interference), alongside a wider spread of CMI classifications across the remaining categories. Taken together, these classifications reflect how participants adjust their allocation of attention in response to changing task demands. The variability in interference indicates that attentional resources are not fixed but flexibly distributed according to contextual demands. This concept aligns with established attentional frameworks.

Among this, multiple resource theory suggests that interference can be reduced in two ways: 1) when tasks draw from separate attention pools, and 2) when the overall demand remains manageable[187,251]. In this case, the verbal working-memory task and visuomotor cutting likely drew from distinct processing resources, allowing participants to maintain or even improve cognitive performance without compromising motor execution. Moreover, for some individuals, the combined tasks likely remained within manageable limits, minimizing interference. In contrast, central capacity theory posits a single, limited pool of attentional resources that must be shared among tasks[171,186,252]. When overall demands exceed this capacity, such as when cognitive and motor tasks are both challenging, resources are reallocated, often favoring motor execution at the expense of cognitive accuracy[252]. This redistribution is likely to account for motor-related cognitive interference, where digit recall accuracy declined while motor performance was reserved, particularly seen in the unanticipated condition. Extending beyond fixed-capacity models, the Executive-Process/Interactive Control (EPIC) framework emphasizes individual differences in the allocation and coordination of managing attention across simultaneous tasks[181,253]. It suggests that multiple tasks can be performed in parallel when the underlying control processes are well-practiced, offering a rationale for the variability observed in CMI patterns across participants [254]. Individual differences in experience, task familiarity, and the degree to which the run-and-cut movement was automatic likely influenced how efficiently participants managed concurrent tasks [254,255]. Those with refined or automatic motor control may have required fewer attentional resources to plan and execute the cut, sustaining cognitive performance. Conversely, participants who relied on more deliberate control likely experienced cognitive-motor interference, stemming from the need to consciously monitor foot placement, speed, and direction change. Such variability in control strategy may also occur within individuals across anticipated and unanticipated conditions, reflecting adaptive attention shifts with changing task demands. Taken together, the CMI classification patterns and theoretical perspectives highlight that dual-task performance reflects not a uniform response, but a continuum of cognitive–motor trade-offs. This underscores the importance of interpreting ACL injury–related mechanics within the broader context of attentional resource allocation and adaptive control.

The observed strategies encompassed facilitation, interference, or no change, emphasizing that the variability itself carries meaningful information about attentional allocation under dual-task conditions. This notion aligns with prior works showing that cognitive load can modify neuromuscular coordination and attentional focus during dynamic movement tasks [244,245]. Such patterns suggest that even minor or non-significant deviations in joint mechanics may reflect adaptive trade-offs in attention rather than uniform performance changes. When attentional resources are directed toward maintaining cognitive performance, the motor system may default to more constrained strategies, consistent with central capacity and resource-sharing models. From a practical perspective, these results underscore the importance of promoting automaticity in sport-specific movements so that attentional resources can be flexibly allocated under cognitive load.

In sport settings, dual cognitive-motor demands are constant, requiring athletes to regulate attention while executing motor control [240,245]. Understanding the range of strategies adopted under such conditions can inform rehabilitation and multifaceted training paradigms particularly prior to returning to play [20,207,239]. Enhancing movement automaticity through practice or task-specific training may reduce interference and mitigate mechanical patterns associated with elevated injury risk.

Future work should examine whether distinct CMI profiles are associated with or predictive of injury risk. This is particularly important in the context of ACL injury, as the distribution of cognitive demand is further influenced by factors such as arousal, individual dispositions, task novelty, and environmental context [256]. Attention focus may also play a critical role, as adopting an external focus of attention has been shown to enhance automaticity and reduce conscious control demands, potentially mitigating interference effects during sport movements [257].

To advance this line of research, several limitations must be acknowledged. The modest sample size may have limited statistical power to detect subtle between-condition effects. Additionally, the cognitive task could have produced ceiling effects that constrained sensitivity to performance differences. Furthermore, the use of controlled, laboratory-based cutting tasks may not fully replicate the perceptual uncertainty and reactive demands of sport competition [232].

Conclusion

Overall, this study showed that dual-task performance during run-and-cut movements represents a spectrum of cognitive–motor strategies rather than a uniform response. While group-level biomechanics were unchanged, variations in attentional allocation suggested subtle adjustments in movement control consistent with shared-resource models of attention. These findings emphasize that cognitive load can influence movement organization even when overt performance appears stable, highlighting the need to incorporate cognitive demands into biomechanical assessment and sport-specific training.

Table 4. Group mean (± SD) knee joint angles (°) at initial contact and global peak. NOCOG_AN represents the baseline motor single-task condition. Dual-task conditions include COG_AN (cognitive anticipated) and COG_UN (cognitive unanticipated).

		NOCOG_AN	COG_AN	COG_UN
Initial Contact	Flexion/Extension	-28.62 ± 9.65	-27.68 ± 8.76	-40.43 ± 16.01
	Abduction/Adduction	-1.01 ± 3.75	-1.01 ± 3.75	-2.11 ± 3.56
	External/Internal Rotation	-1.85 ± 7.95	-1.72 ± 6.19	3.32 ± 8.20
Peak	Flexion/Extension	-58.57 ± 7.06	-57.32 ± 5.82	-60.13 ± 9.90
	Abduction/Adduction	-8.06 ± 5.03	-7.72 ± 3.90	-7.99 ± 3.67
	External/Internal Rotation	13.11 ± 4.81	13.41 ± 3.21	13.17 ± 4.83

Table 5. Group mean (± SD) peak knee joint moments (Nm/kg). NOCOG_AN represents the baseline motor single-task condition. Dual-task conditions include COG_AN (cognitive anticipated) and COG_UN (cognitive unanticipated).

	NOCOG_AN	**COG_AN**	**COG_UN**
Extension	3.42 ± 0.53	3.31 ± 0.61	2.87 ±0.64
Adduction	0.50 ±0.30	0.48 ± 0.31	0.41 ± 0.31
External Rotation	-0.18 ± 0.13	-0.17 ± 0.15	-0.24 ± 0.67

Table 6. Group mean (± SD) peak GRF (BW). NOCOG_AN represents the baseline motor single-task condition. Dual-task conditions include COG_AN (cognitive anticipated) and COG_UN (cognitive unanticipated).

	NOCOG_AN	COG_AN	COG_UN
Vertical GRF	2.28 ± 0.26	2.27 ± 0.27	2.14 ± 0.30
Anteroposterior GRF	-0.74 ± 0.20	-0.68 ± 0.21	-0.49 ± 0.36
Mediolateral GRF	0.56 ± 0.77	0.59 ± 0.76	0.56 ± 0.72

Table 7. Grouped cognitive–motor interference (CMI) outcomes by number and percentage of participants (n, %) for ACL-injury relevant mechanics.

	CMI Classification Outcome	
	Anticipated	**Unanticipated**
Peak Knee Flexion Angle	Cognitive facilitation 7 (41.2) Motor-related cognitive interference 5 (29.4) No dual-task interference 5 (29.4)	Cognitive facilitation 4 (23.5) Motor-related cognitive interference 5 (29.4) No dual-task interference 6 (35.0) Motor-priority trade-off 1 (5.8) Motor facilitation 1 (5.8)
Peak Knee Abduction Angle	Cognitive facilitation 7 (41.2) Motor-related cognitive interference 5 (29.4) No dual-task interference 5 (29.4)	Cognitive facilitation 4 (23.5) Motor-related cognitive interference 6 (35.3) No dual-task interference 7 (41.2)
Peak Knee Adduction Moment	Cognitive facilitation 7 (41.2) Motor-related cognitive interference 5 (29.4) No dual-task interference 5 (29.4)	Cognitive facilitation 4 (23.5) Motor-related cognitive interference 6 (35.3) No dual-task interference 6 (35.3) Motor facilitation 1 (5.8)

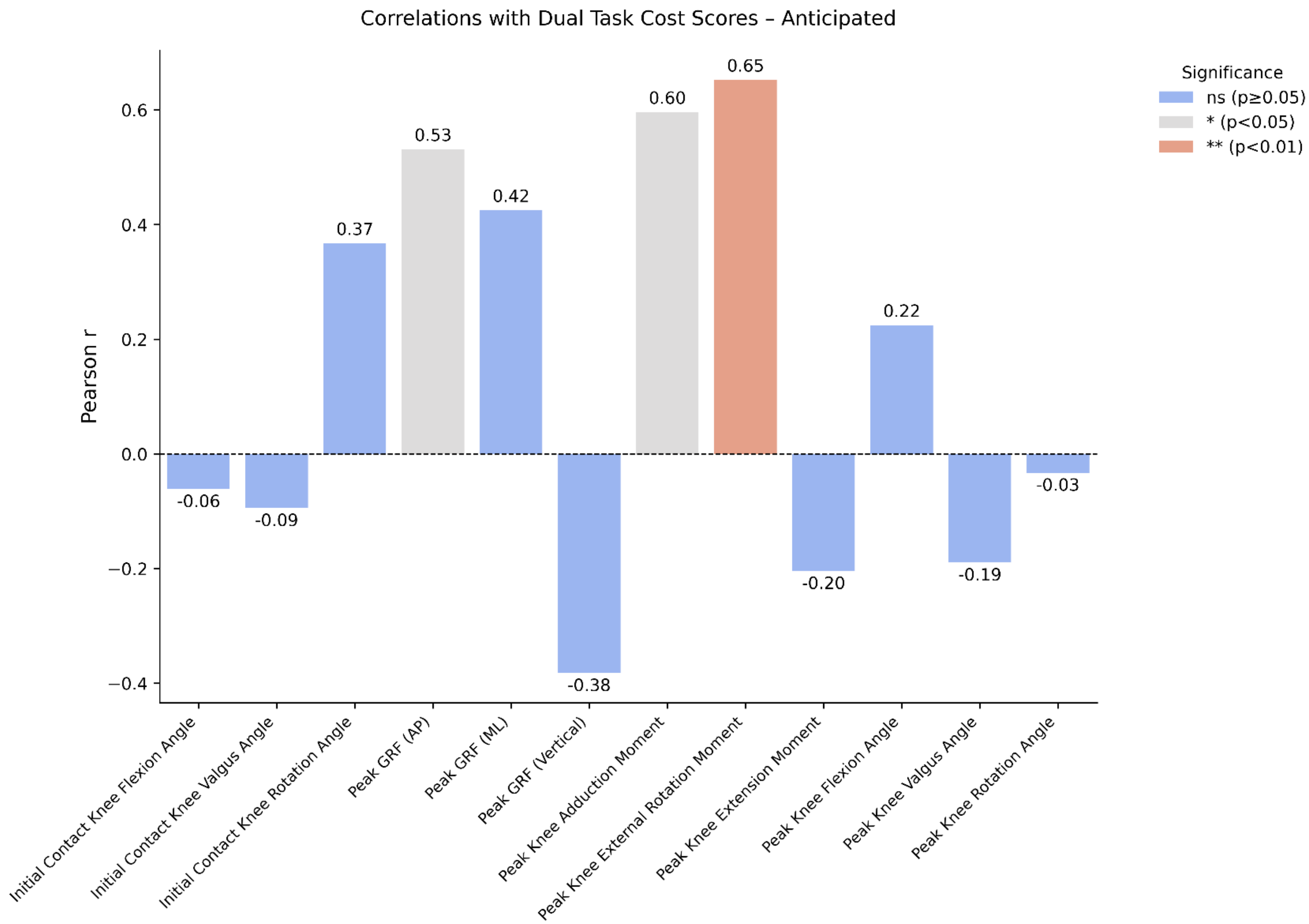

Figure 6. Correlations between pooled dual task cost scores and knee joint mechanic variables under anticipated conditions

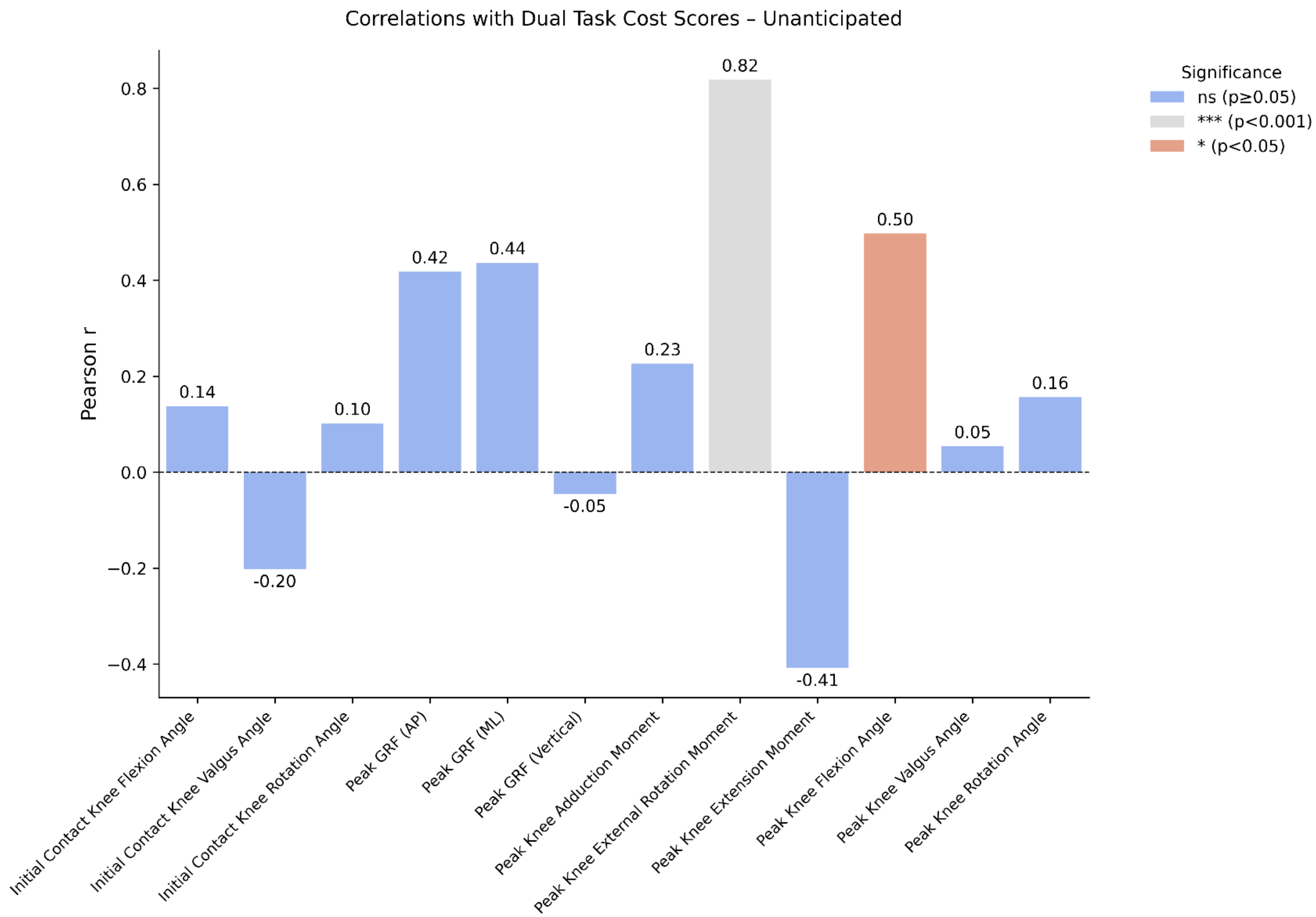

Figure 7. Correlations between pooled dual task cost scores and knee joint mechanic variables under unanticipated conditions

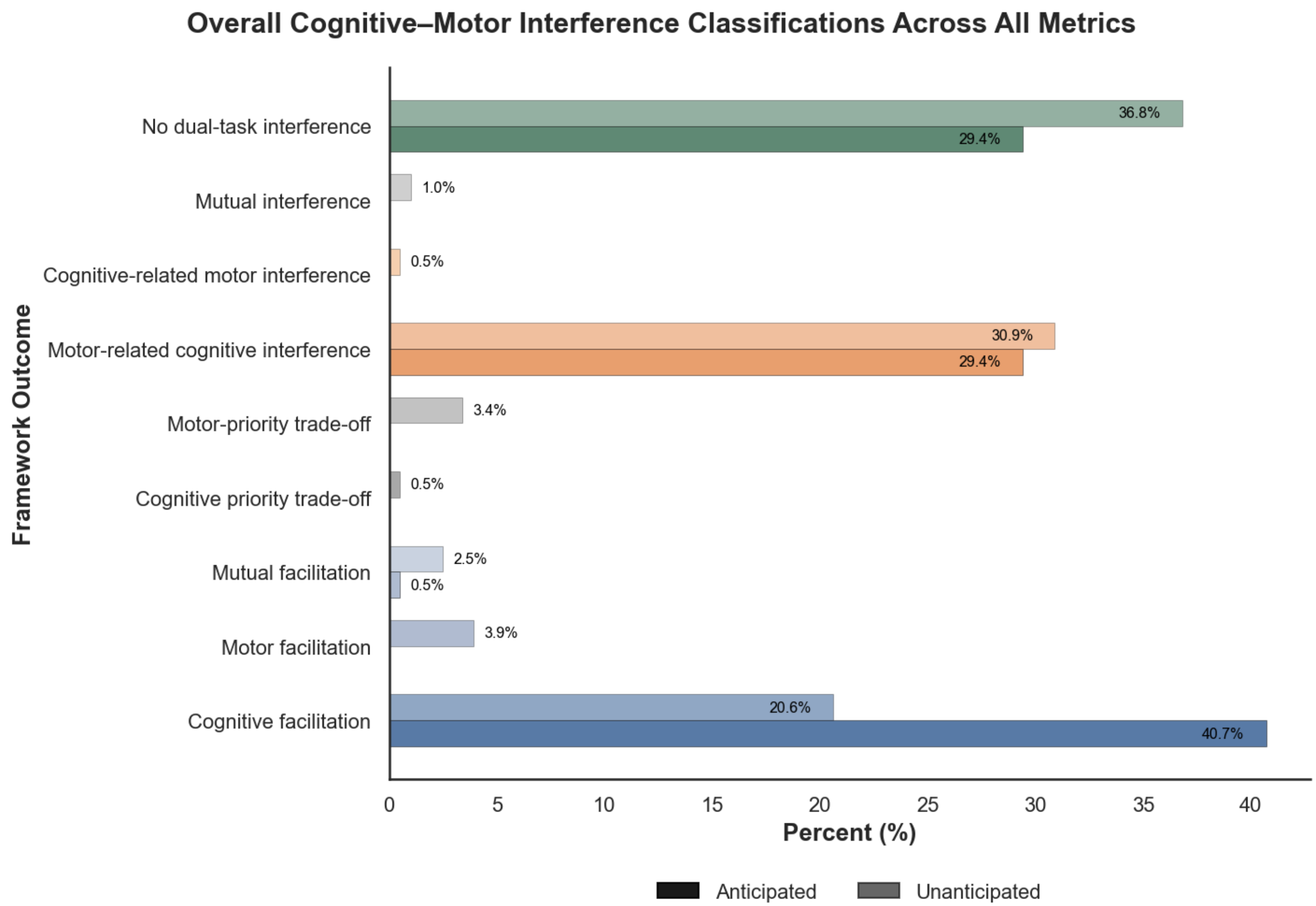

Figure 8. Distribution of pooled cognitive-motor interference (CMI) classifications across all variables and participants. Percentages represent the proportion of all metric-level classifications summed across participants for anticipated and unanticipated conditions.

Chapter 6: Conclusion

This dissertation sought to examine how cognitive and anticipatory demands influence knee joint mechanics and neuromuscular control during run-and-cut tasks. The overarching goal was to understand how attention is distributed across cognitive-motor dual task demands and how this interaction influences movement strategies associated with ACL injury risk mechanics. Chapter 4 focused on biomechanical and neuromuscular outcomes, presenting anticipation effects, with unanticipated cuts exhibiting greater knee flexion angles. Cognitive load produced subtle effects, suggesting reallocation of attentional resources under dual task conditions. No significant interaction between anticipation and cognitive load were found, suggesting independent effects. However, at the group level, statistical analysis masked individual mechanisms.

Chapter 5 extended this work by classifying patterns of CMI based on quantitatively measuring changes in motor and cognitive performance. These results demonstrated that participants managed dual-task demands differently, with some maintaining or even enhancing cognitive performance without measurable decrements in motor execution, while others exhibited declines consistent with cognitive- or motor-related interference. This individualized approach emphasizes that variability in CMI reflects strategic allocation of attention rather than uniform degradation of motor performance.

Collectively, these findings provide converging evidence that cognitive load and anticipation shape movement behavior through distinct yet interacting attentional pathways. By integrating biomechanical, neuromuscular, and cognitive data, this work advances a more comprehensive understanding of how cognitive-motor processes contribute to knee joint control and ACL injury risk.

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Appendices

Appendix A. Subject Characteristics

Table 8. Individual subject characteristics.

Participant	Age (yrs)	Height (m)	Mass (kg)	Dominant Limb
S01	26	1.65	50.97	R
S03	21	1.63	67.28	R
S04	22	1.70	60.45	R
S05	22	1.63	55.05	R
S06	20	1.70	67.90	R
S07	28	1.70	79.51	R
S08	26	1.68	57.70	R
S09	19	1.73	50.97	R
S10	21	1.63	55.86	L
S11	26	1.78	65.34	R
S12	23	1.69	72.48	R
S13	21	1.64	63.71	R
S14	18	1.69	61.57	R
S15	21	1.77	69.72	R
S16	20	1.66	62.28	L
S17	22	1.73	63.00	L
S18	24	1.65	66.56	R
Mean ± SD	22.4 ± 2.8	1.69 ± 0.05	62.9 ± 7.5	

Table 9. Subject-level means of initial contact knee flexion/extension angle across all conditions. Each value represents the mean across all successful trials within each subject-condition. Negative values indicate knee flexion, and positive values indicate knee extension.

	Initial Contact Knee Flexion/Extension (°)			
Subject	NOCOG_AN	NOCOG_UN	COG_AN	COG_UN
S01	-27.2	-35.3	-28.1	-38.1
S03	-29.4	-43.8	-27.4	-39.5
S04	-33.7	-56.8	-30.9	-47.9
S05	-25.1	-40.4	-27.7	-44.8
S06	-27.5	-53.4	-23.8	-54.4
S07	-44.5	-48.8	-42.1	-38.9
S08	-26	-47	-24.5	-45.9
S09	-42.3	-56.8	-36.9	-56.1
S10	-28	-28	-25.5	-48.3
S11	-42	-62.9	-32.8	-61.7
S12	-30.5	-57.8	-22.1	-47.1
S13	-25.9	-42.8	-25.8	-39.7
S14	-15.6	-16.9	-17.7	-18.8
S15	-29.7	-41.6	-32.3	-27.4
S16	-22.9	-25.5	-16.5	-21.8
S17	-18.1	-28.4	-31	-25.5
S18	-19	-34	-18.6	-36.2
Mean ± SD	-28.7 ± 8.2	42.4 ± 13.0	-27.3 ± 6.8	-40.7 ± 12.1

Table 10. Subject-level means of initial contact knee abduction/adduction angle across all conditions. Each value represents the mean across all successful trials within each subject-condition. Negative values indicate abduction, and positive values indicate adduction.

	Initial Contact Knee Abduction/Adduction (°)			
Subject	NOCOG_AN	NOCOG_UN	COG_AN	COG_UN
S01	2.7	2.5	2	1.3
S03	-3.4	-6.7	-3.5	-4.6
S04	-3.1	-2.6	-4.3	-1.9
S05	-4	-2.6	-2.3	-3.4
S06	-5.2	-3.9	-4.5	-0.8
S07	-0.9	-1.7	0.3	-2.1
S08	0.7	-3	-1.1	-4.6
S09	1.8	0.5	2.3	-4
S10	3.1	-0.8	3.3	3
S11	8.1	6.7	7.3	2.7
S12	-2.1	-3.6	-2.4	-3.8
S13	-0.7	-2.9	-2.2	-2.1
S14	-0.3	-0.3	-0.8	0.1
S15	-4.8	-5.6	-5.7	-3.8
S16	-1.3	-1.6	-0.1	-3.2
S17	-2.7	-1.3	-1.2	-2.1
S18	-4.8	-4.8	-6.2	-7.4
Mean ± SD	-1.0 ± 3.5	-1.9 ± 3.1	-1.1 ± 3.5	-2.2 ± 2.7

Table 11. Subject-level means of initial contact knee external/internal rotation angle across all conditions. Each value represents the mean across all successful trials within each subject-condition. Positive values indicate internal rotation, and negative values indicate external rotation.

	Initial Contact Knee External/Internal Rotation Angle (°)			
Subject	NOCOG_AN	NOCOG_UN	COG_AN	COG_UN
S01	-0.4	4.3	0.1	6.1
S03	-5.5	-1.3	0.2	2.6
S04	-2	2.9	-1.8	-1.3
S05	2.5	9.6	0.3	14.5
S06	1.5	2.9	0.3	-0.2
S07	1.8	2.5	2.2	7.2
S08	-3.8	9.1	1.4	12.7
S09	9.2	7.9	5.2	15
S10	-13.7	-7.3	-15.1	-3.9
S11	-5.8	4.3	-8.4	3.4
S12	7.2	6.8	1.8	2.5
S13	-6.3	-4	-3.4	-3.2
S14	-12.2	-10.2	-10.1	-10.7
S15	-4.7	2.5	-2.9	-9.3
S16	3.8	5.7	-2	9.4
S17	0.8	4.6	-0.1	3.1
S18	-3.9	-2.6	0.9	7.1
Mean ± SD	-1.9 ± 6.1	2.2 ± 5.6	-1.8 ± 5.0	3.2 ± 7.5

Table 12. Subject-level means of peak knee extension/flexion angle across all conditions. Each value represents the mean across all successful trials within each subject-condition. Negative values indicate knee flexion, and positive values indicate knee extension.

	Peak Knee Extension/Flexion Rotation Angle (°)			
Subject	NOCOG_AN	NOCOG_UN	COG_AN	COG_UN
S01	-59.2	-56	-59.4	-59.4
S03	-55.9	-62.6	-55.2	-59
S04	-49.2	-49.4	-45.2	-44.8
S05	-60	-52.8	-58.1	-54
S06	-61.7	-69.6	-61.3	-69.4
S07	-65.8	-66.3	-62.9	-62.8
S08	-56.7	-58.5	-53.6	-54.5
S09	-68.7	-73.7	-65.8	-72.8
S10	-65.7	-66	-64.7	-78.6
S11	-60.7	-67.6	-56.5	-65.4
S12	-62.7	-64	-60.4	-63.5
S13	-64.9	-66	-64.8	-66.3
S14	-48.6	-52.5	-51.8	-53.1
S15	-51.5	-52.6	-55.9	-65.8
S16	-56	-53.1	-54.6	-56.9
S17	-52.9	-55.8	-53.2	-53.2
S18	-52.5	-52.5	-52.2	-48.3
Mean ± SD	-58.4 ± 6.1	-59.9 ± 7.4	57.4 ± 5.5	-60.5 ± 8.9

Table 13. Subject-level means of peak knee abduction/adduction angle across all conditions. Each value represents the mean across all successful trials within each subject-condition. Negative values indicate abduction, and positive values indicate adduction.

	Peak Knee Abduction/Adduction Angle (°)			
Subject	NOCOG_AN	NOCOG_UN	COG_AN	COG_UN
S01	-2.4	-3.3	-3	-3.5
S03	-7.8	-8.2	-8.5	-7.3
S04	-8.6	-7.1	-7.4	-6.1
S05	-11.2	-7.9	-9.7	-7.9
S06	-12.4	-9.2	-11.2	-7.4
S07	-7.1	-6.4	-7.1	-6.2
S08	-5.5	-8.6	-5.3	-6.8
S09	-6	-6.1	-3.1	-10.6
S10	-2.9	-6.3	-3.9	-7.3
S11	-0.6	1.8	-2	-0.2
S12	-19.6	-16.9	-12	-12.1
S13	-13.2	-13.8	-14	-11.7
S14	-3	-6.4	-5.6	-6.9
S15	-9.4	-10.8	-13.3	-14
S16	-11.2	-12.4	-9.6	-13.3
S17	-5.1	-10.2	-7.9	-9.9
S18	-8.4	-8.6	-11.6	-7
Mean ± SD	-7.9 ± 4.7	-8.3 ± 4.2	-8.0 ± 3.8	-8.1 ± 3.5

Table 14. Subject-level means of peak knee external/internal rotation angle across all conditions. Each value represents the mean across all successful trials within each subject-condition. Positive values indicate internal rotation, and negative values indicate external rotation.

	Peak Knee External/Internal Rotation Angle (°)			
Subject	NOCOG_AN	NOCOG_UN	COG_AN	COG_UN
S01	12.3	13.8	13.2	14.4
S03	10.2	10.3	11.2	10.1
S04	16.2	13.3	14.4	12.3
S05	15.8	15	14.6	16.3
S06	9.1	6.5	9.7	4.7
S07	14.9	12.7	13.6	11.5
S08	15.6	18.3	15.2	17.2
S09	21.5	19.5	17.9	22
S10	6.6	9.5	10.5	8.7
S11	12.4	11.3	12.2	10.5
S12	22.3	18.9	18	15.9
S13	13.1	14.8	13.3	14.9
S14	6.2	6	7.7	7.6
S15	7.9	6.7	10.4	9.6
S16	14.1	14.6	12.8	15.4
S17	15.9	21.1	19.1	20.2
S18	10.7	11.1	13.5	11.9
Mean ± SD	13.2 ± 4.6	13.1 ± 4.6	13.4 ± 3.1	13.1 ± 4.5

Table 15. Subject-level means of peak knee extension moment across all conditions. Each value represents the mean across all successful trials within each subject-condition.

	Peak Knee Extension Moment (Nm/kg)			
Subject	NOCOG_AN	NOCOG_UN	COG_AN	COG_UN
S01	3.3	2.8	3	2.7
S03	2.9	2.1	2.8	2.4
S04	3.7	2.6	3.8	2.9
S05	3.1	2.3	3.2	2.2
S06	3	2.5	3.2	2.4
S07	3.6	3.2	3.3	3
S08	3.9	3.1	4	2.8
S09	3.1	2	2.9	2.4
S10	4	3.9	4.5	4
S11	4.1	2.9	4	2.8
S12	3.6	3.1	3.8	3.1
S13	3.1	3	3.4	3.2
S14	3.8	3.4	3.3	3.3
S15	3.4	2.2	3.4	4
S16	3.6	2.7	3	3.3
S17	2.5	2.1	2.2	2.3
S18	2.9	2.4	2.8	2.3
Mean ± SD	3.4 ± 0.4	2.7 ± 0.5	3.3 ± 0.6	2.9 ± 0.6

Table 16. Subject-level means of peak knee adduction moment across all conditions. Each value represents the mean across all successful trials within each subject-condition.

	Peak Knee Adduction Moment (Nm/kg)			
Subject	NOCOG_AN	NOCOG_UN	COG_AN	COG_UN
S01	0.7	0.1	0.8	0.1
S03	0.1	0.1	0.1	0.2
S04	0.2	0.1	0.3	0.3
S05	0.7	0.3	0.5	0.3
S06	0.5	0.4	0.3	0.2
S07	0.6	0.5	0.8	0.5
S08	0.2	0.3	0.4	0.5
S09	0.5	0.5	0.6	0.7
S10	0.2	0.2	0.1	0.1
S11	0.4	0.2	0.5	0.2
S12	0.9	0.7	0.7	0.6
S13	0.5	0.2	0.4	0.1
S14	0.8	0.8	0.7	0.8
S15	0.9	1.4	1	1.1
S16	0.6	0.4	0.6	0.6
S17	0.4	0.3	0.2	0.3
S18	0.2	0.1	0.3	0.1
Mean ± SD	0.5 ± 0.3	0.4 ± 0.3	0.5 ± 0.3	0.4 ± 0.3

Table 17. Subject-level means of peak knee external rotation moment across all conditions. Each value represents the mean across all successful trials within each subject-condition.

	Peak Knee External Rotation Moment (Nm/kg)			
Subject	NOCOG_AN	NOCOG_UN	COG_AN	COG_UN
S01	-0.2	-0.2	-0.2	-0.3
S03	-0.1	0.0	0.0	-0.1
S04	-0.2	-0.2	-0.1	-0.1
S05	-0.1	-0.2	-0.1	-0.1
S06	-0.1	-0.1	-0.1	0.0
S07	-0.3	-0.2	-0.2	-0.1
S08	-0.1	-0.1	0.0	-0.1
S09	-0.3	-0.2	-0.3	-0.2
S10	-0.4	-0.2	-0.6	-2.0
S11	-0.3	-0.1	-0.3	-0.2
S12	-0.2	-0.2	-0.1	-0.1
S13	-0.1	-0.3	-0.1	-0.3
S14	-0.1	-0.1	-0.1	-0.1
S15	-0.1	0.0	0.0	0.0
S16	-0.2	-0.3	-0.1	-0.2
S17	-0.2	-0.3	-0.3	-0.3
S18	0.0	0.0	0.0	-0.1
Mean ± SD	-0.2 ± 0.1	-0.2 ± 0.1	-0.2 ± 0.2	-0.3 ± 0.5

Table 18. Subject-level means of peak vertical ground reaction force across all conditions. Each value represents the mean across all successful trials within each subject-condition.

	Peak Vertical Ground Reaction Force (BW)			
Subject	NOCOG_AN	NOCOG_UN	COG_AN	COG_UN
S01	2.1	2	1.9	1.9
S03	2.4	1.9	2.2	2.1
S04	2.3	2	2.6	2.3
S05	2.3	2.5	2.5	2.4
S06	2.3	1.9	2.4	2
S07	2.2	2.1	2.1	2.1
S08	2.3	2.1	2.5	2.1
S09	2.2	1.8	2.2	1.8
S10	2.6	2.3	2.8	2.3
S11	2.2	1.8	2.2	1.8
S12	2.1	1.9	2	1.7
S13	2	2.1	2.1	2.1
S14	2.4	2.2	2.3	2.3
S15	2.3	2.1	2.3	2.7
S16	2.6	2.4	2.1	2.5
S17	2.1	2	2.2	2.3
S18	2	1.9	2.1	2.1
Mean ± SD	2.3 ± 0.2	2.1 ± 0.2	2.3 ± 0.2	2.1 ± 0.3

Table 19. Subject-level means of peak anteroposterior ground reaction force across all conditions. Each value represents the mean across all successful trials within each subject-condition. Negative values indicate braking.

Peak Anteroposterior Ground Reaction (BW)				
Subject	NOCOG_AN	NOCOG_UN	COG_AN	COG_UN
S01	-0.8	-0.3	-0.7	-0.4
S03	-0.5	-0.1	-0.4	-0.1
S04	-0.7	-0.2	-0.6	-0.2
S05	-0.7	-0.3	-0.6	-0.3
S06	-0.5	-0.2	-0.6	-0.2
S07	-0.8	-0.5	-0.6	-0.5
S08	-0.7	-0.3	-0.7	-0.3
S09	-0.6	-0.2	-0.6	-0.5
S10	-1.1	-0.9	-1.3	-1.3
S11	-0.6	-0.2	-0.6	-0.2
S12	-0.7	-0.3	-0.8	-0.5
S13	-0.6	-0.5	-0.7	-0.4
S14	-1	-0.9	-0.9	-0.9
S15	-0.8	-0.6	-0.8	-1.2
S16	-0.9	-0.6	-0.7	-0.8
S17	-0.8	-0.5	-0.6	-0.6
S18	-0.5	-0.3	-0.6	-0.2
Mean ± SD	-0.7 ± 0.2	-0.4 ± 0.2	-0.7 ± 0.2	-0.5 ± 0.4

Table 20. Subject-level means of peak mediolateral ground reaction force across all conditions. Each value represents the mean across all successful trials within each subject-condition.

	Peak Mediolateral Ground Reaction Force (BW)			
Subject	NOCOG_AN	NOCOG_UN	COG_AN	COG_UN
S01	0.9	1.0	0.8	0.9
S03	0.8	0.7	0.8	0.8
S04	0.8	0.6	0.8	0.6
S05	1.0	1.0	1.0	0.9
S06	0.9	0.8	1.0	0.8
S07	0.8	1.0	0.9	1.0
S08	1.1	1.1	1.1	1.1
S09	1.0	0.8	1.0	0.8
S10	-0.9	-0.8	-0.9	-0.7
S11	0.9	0.8	0.8	0.8
S12	0.9	1.0	0.9	0.9
S13	0.9	1.0	1.0	0.9
S14	1.0	1.0	1.0	1.0
S15	0.9	1.0	1.1	0.7
S16	-1.1	-1.0	-1.0	-1.1
S17	-0.9	-1.0	1.0	-1.1
S18	1.0	0.9	0.9	0.9
Mean ± SD	0.6 ± 0.7	0.6 ± 0.7	0.6 ± 0.8	0.5 ± 0.7

Table 21. Subject-level mean entrance and exit velocities (m/s) across task conditions.

	NOCOG_AN		NOCOG_UN		COG_AN		COG_UN	
	Entrance	Exit	Entrance	Exit	Entrance	Exit	Entrance	Exit
S01	3.57	3.45	3.39	3.77	3.61	3.39	4.14	3.57
S03	3.87	3.13	3.63	3.33	3.92	3.06	3.58	2.83
S04	3.70	3.43	3.24	3.21	3.51	3.46	3.56	3.23
S05	3.75	3.09	3.46	2.97	3.72	3.16	3.48	3.13
S06	4.04	3.67	3.76	2.65	3.77	3.75	3.77	3.51
S07	4.14	3.05	3.68	2.73	4.25	3.25	3.83	2.97
S08	3.78	3.70	3.58	3.50	3.76	3.64	3.67	3.43
S09	4.76	4.35	4.26	3.25	4.48	4.17	4.35	3.26
S10	4.04	3.45	3.95	3.37	3.94	3.37	3.82	3.33
S11	3.70	3.57	3.54	3.28	3.44	3.37	3.35	3.41
S12	3.94	3.08	3.70	3.23	3.92	3.28	3.81	3.53
S13	3.70	3.13	3.58	3.18	3.62	3.19	3.31	3.45
S14	3.94	3.10	3.92	2.94	3.82	3.09	3.76	2.80
S15	4.08	3.09	4.43	3.45	4.13	3.49	3.88	3.66
S16	4.24	3.28	3.86	3.13	4.18	3.37	4.10	3.06
S17	4.02	3.37	3.67	3.53	4.04	3.57	3.84	3.74
S18	3.65	--	3.57	--	3.64	--	3.51	--
Mean ± SD	3.94 ± 0.29	3.37 ± 0.34	3.72 ± 0.30	3.22 ± 0.29	3.87 ± 0.28	3.41 ± 0.28	3.75 ± 0.28	3.31 ± 0.28

Appendix C. Individual Subject Ensemble Curves for Knee Joint Kinematics

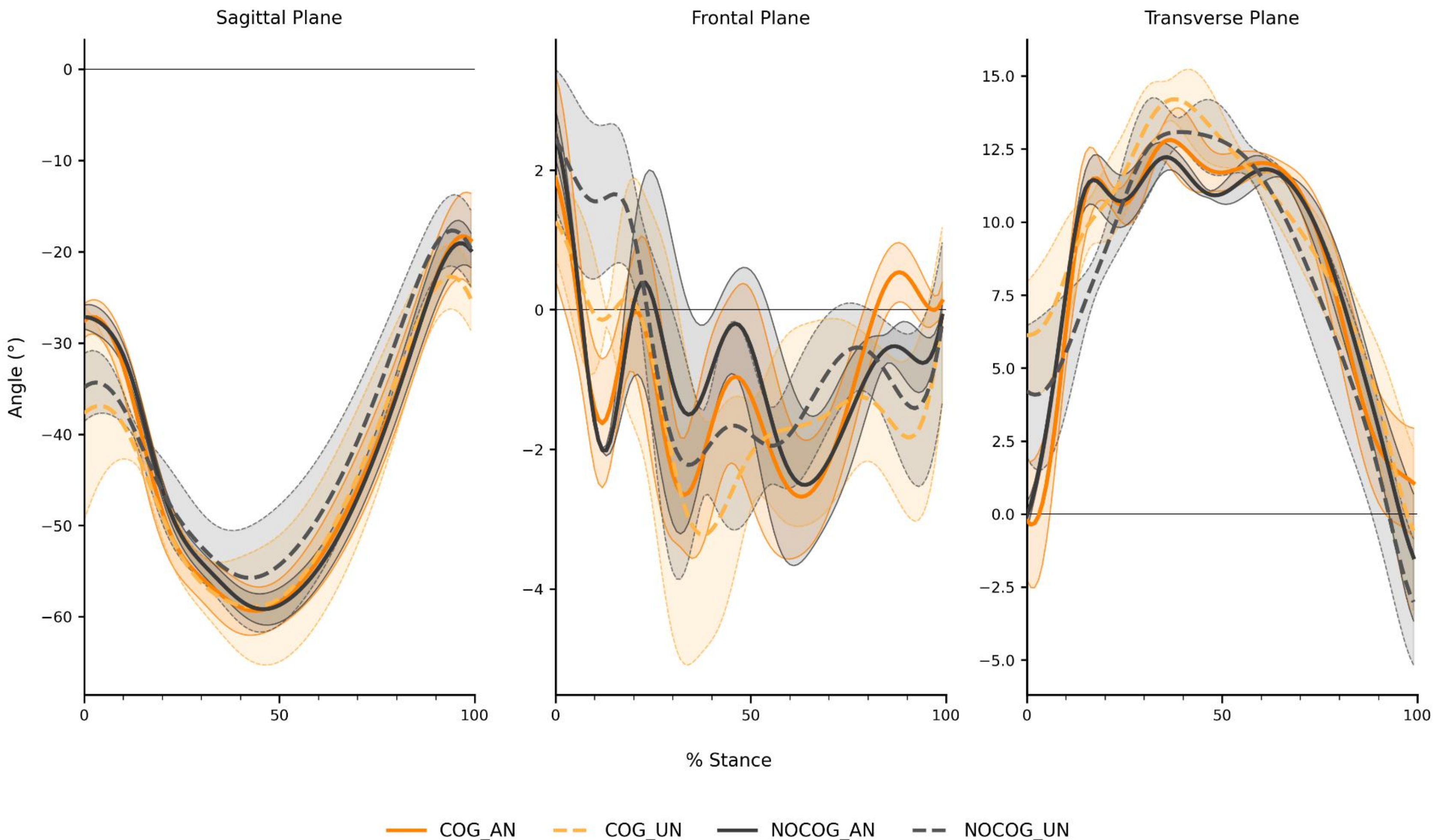

Figure 9. Subject 01 three-dimensional knee joint angles from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee angle for each condition.

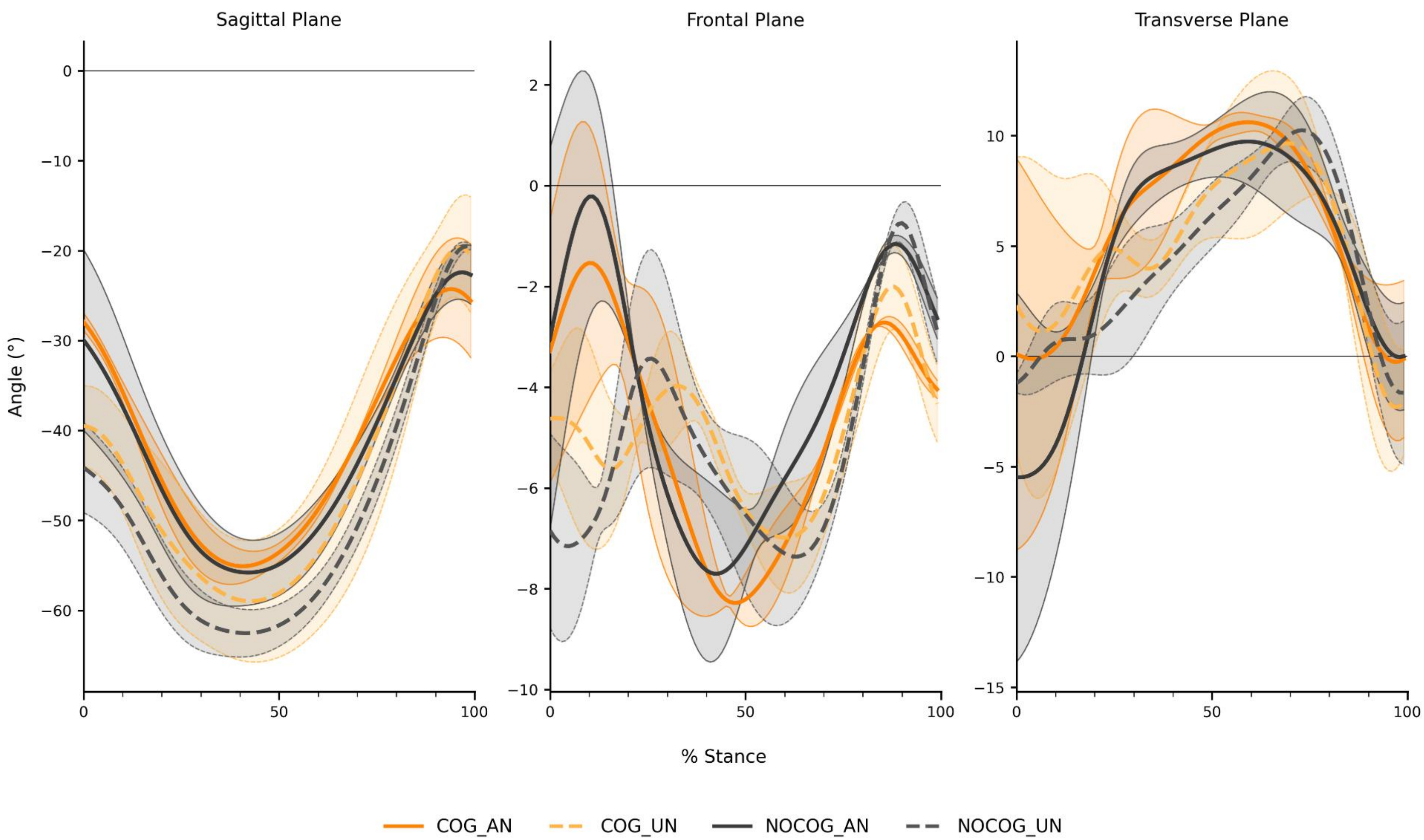

Figure 10. Subject 03 three-dimensional knee joint angles from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee angle for each condition.

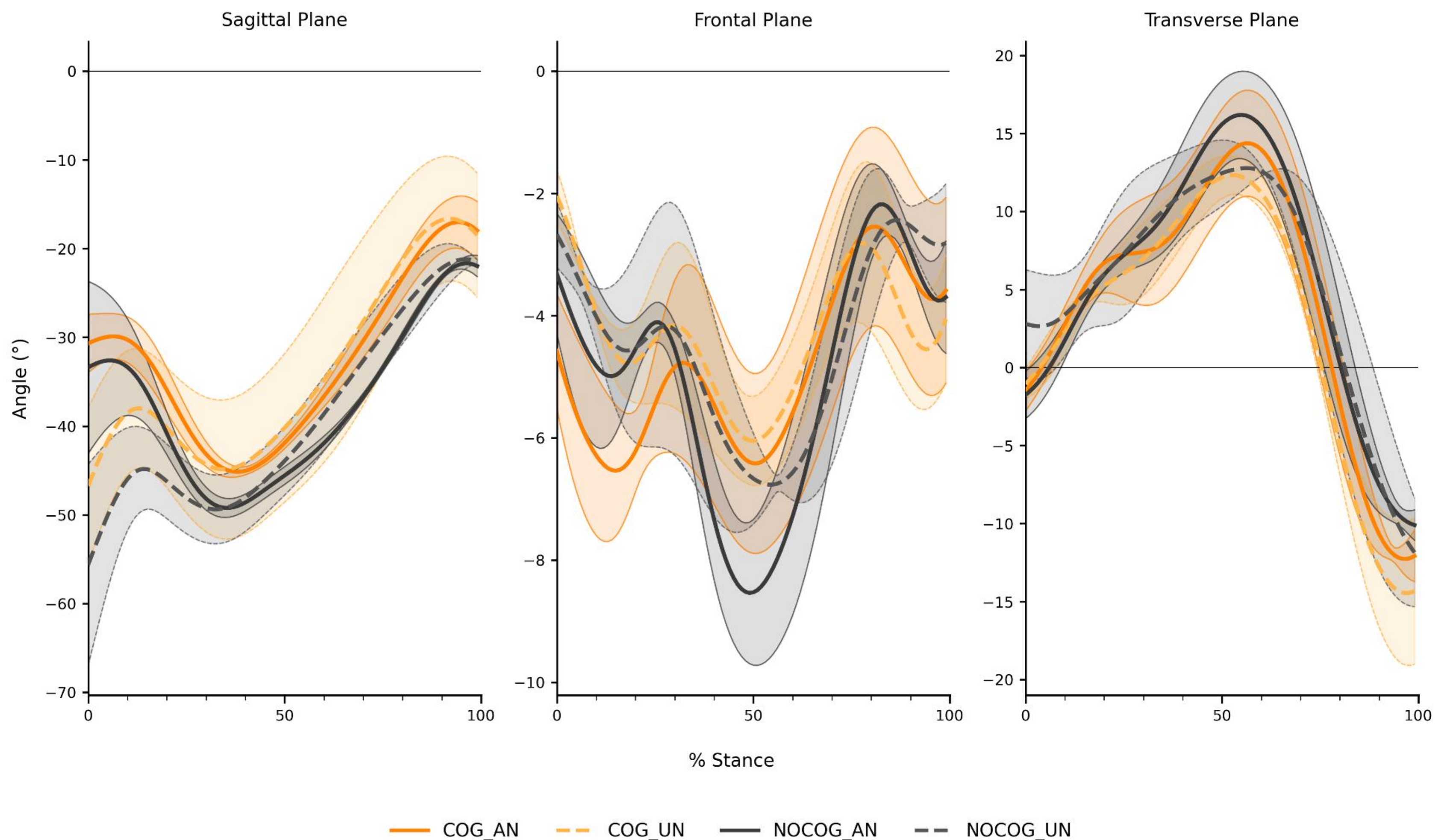

Figure 11. Subject 04 three-dimensional knee joint angles from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee angle for each condition.

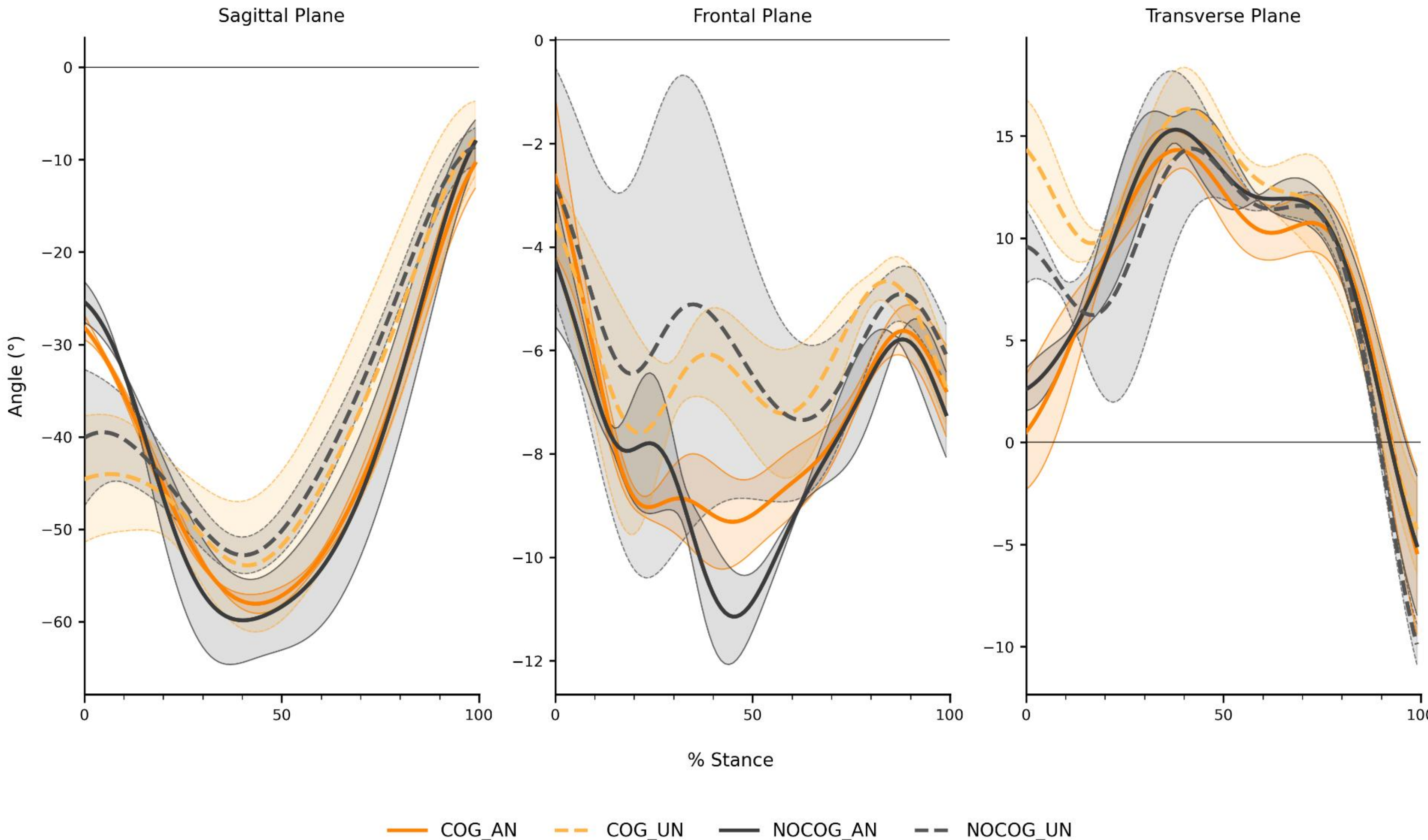

Figure 12. Subject 05 three-dimensional knee joint angles from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee angle for each condition.

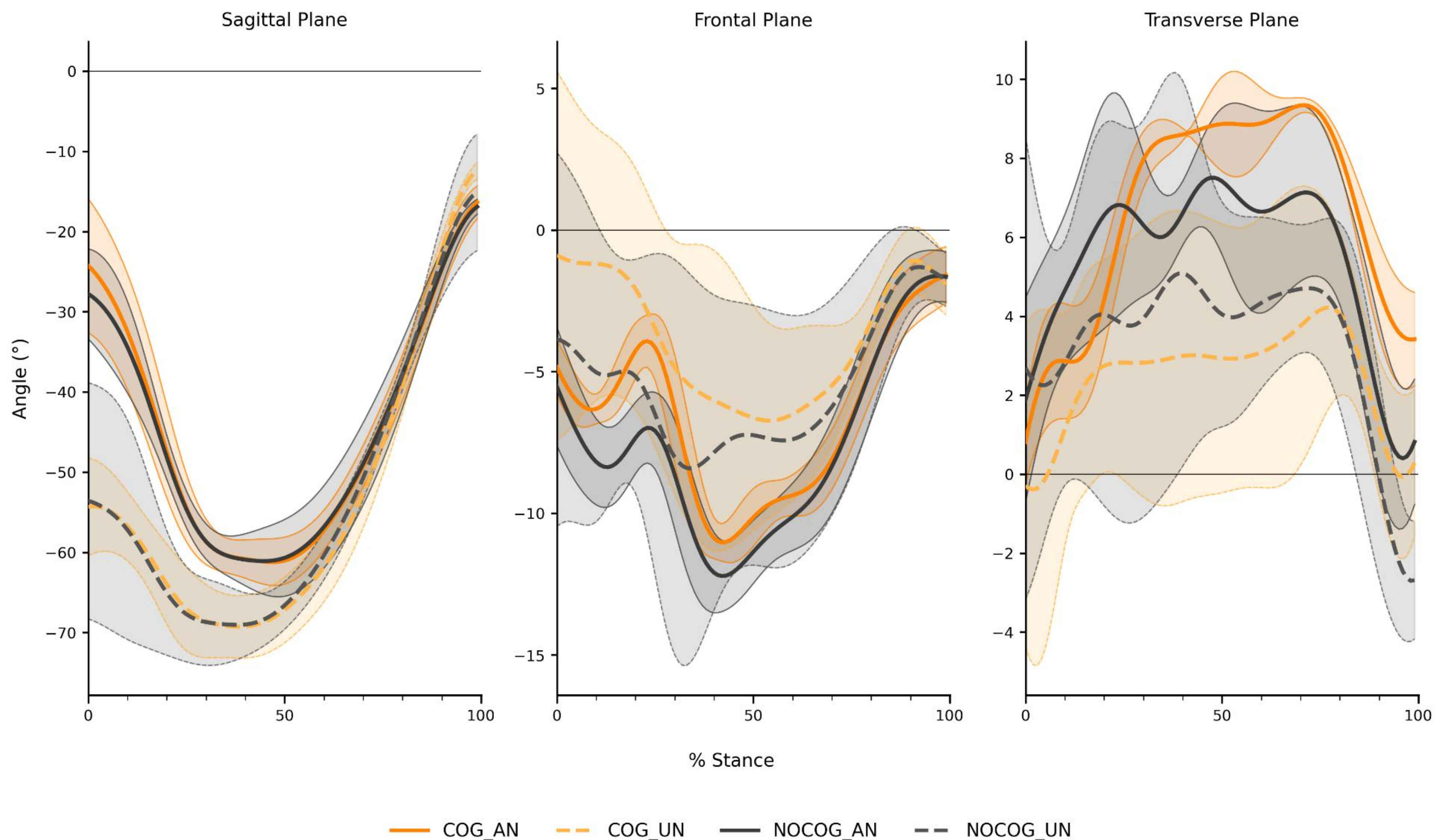

Figure 13. Subject 06 three-dimensional knee joint angles from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee angle for each condition.

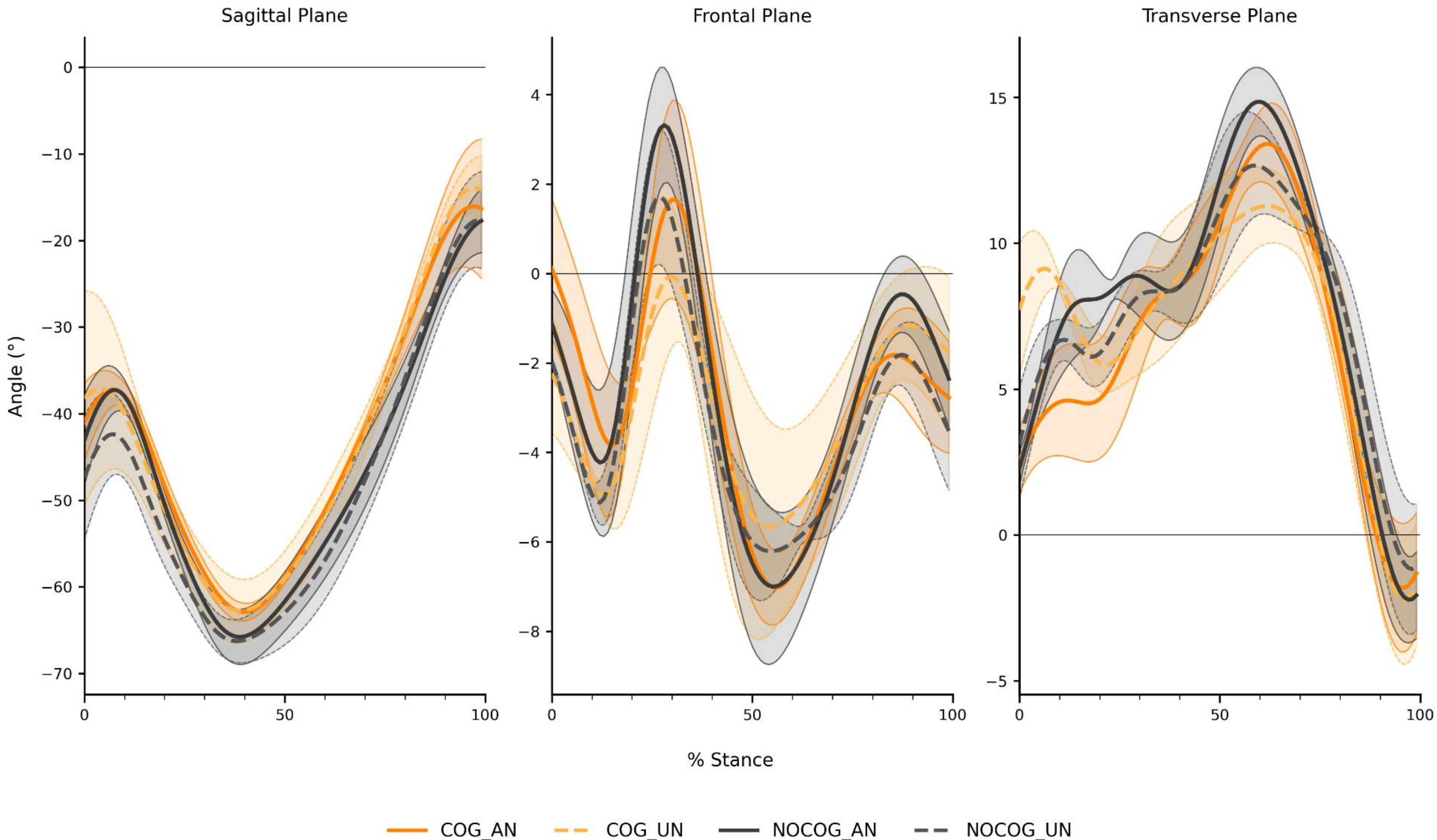

Figure 14. Subject 07 three-dimensional knee joint angles from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee angle for each condition.

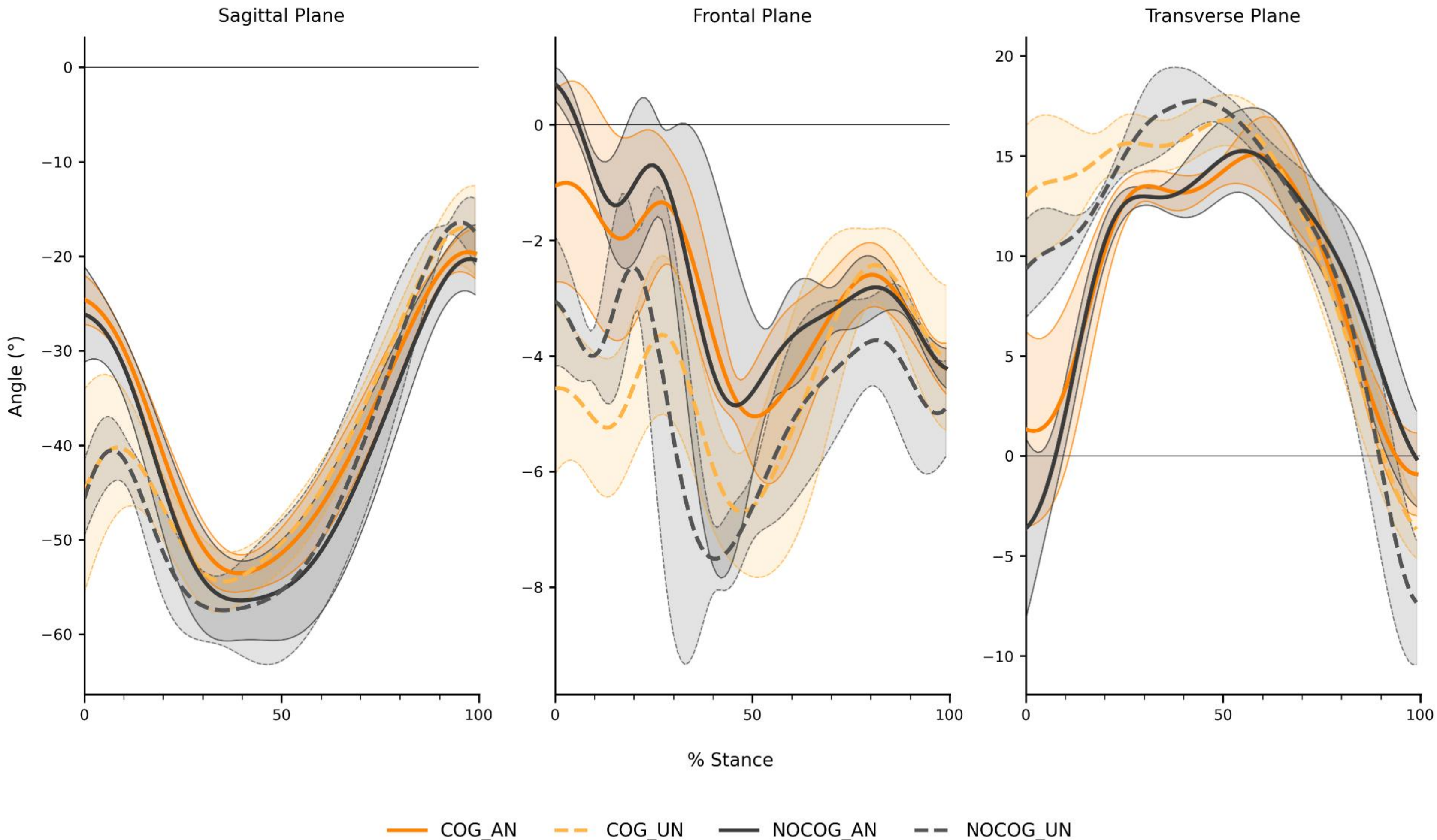

Figure 15. Subject 08 three-dimensional knee joint angles from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee angle for each condition.

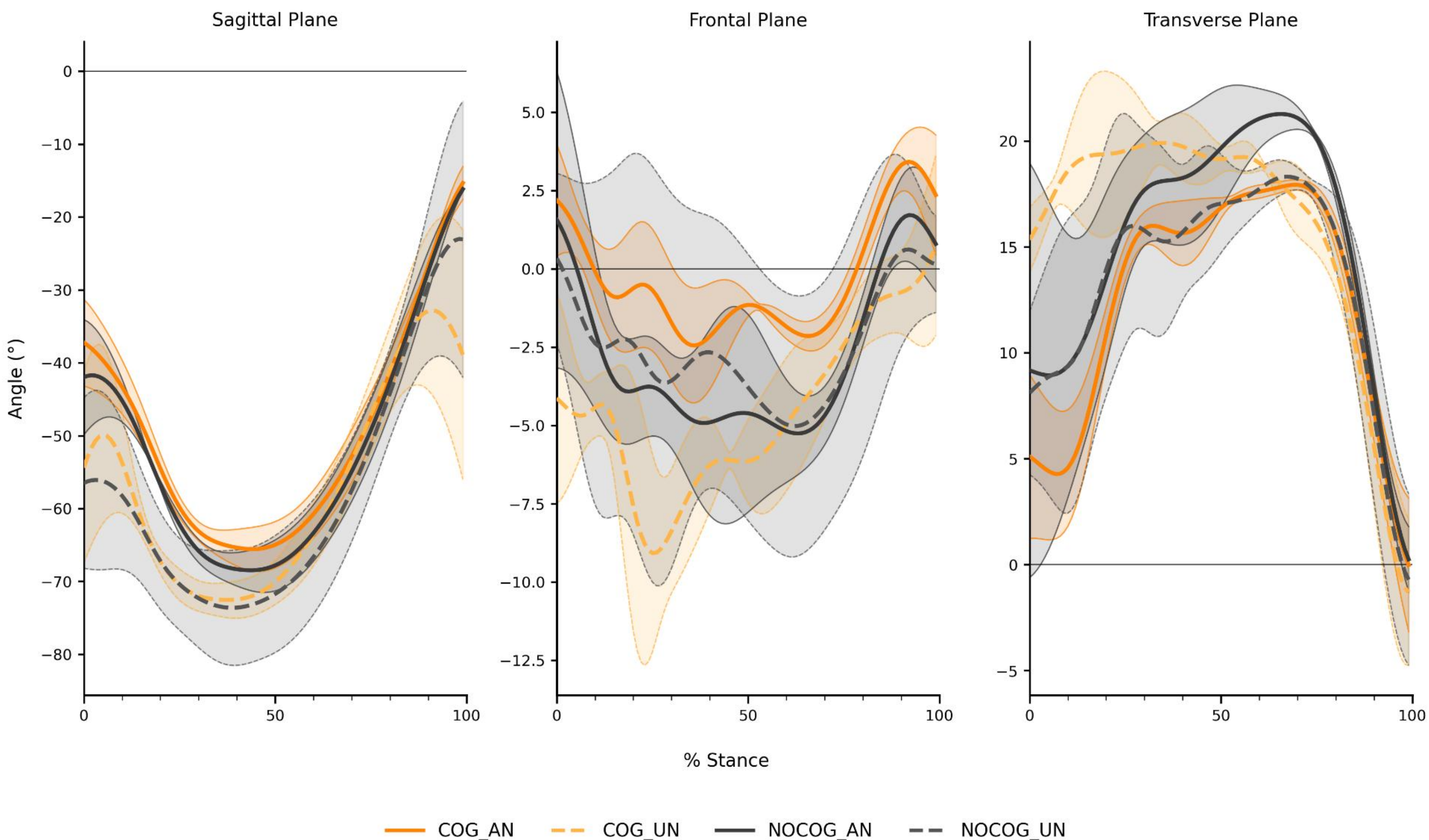

Figure 16. Subject 09 three-dimensional knee joint angles from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee angle for each condition.

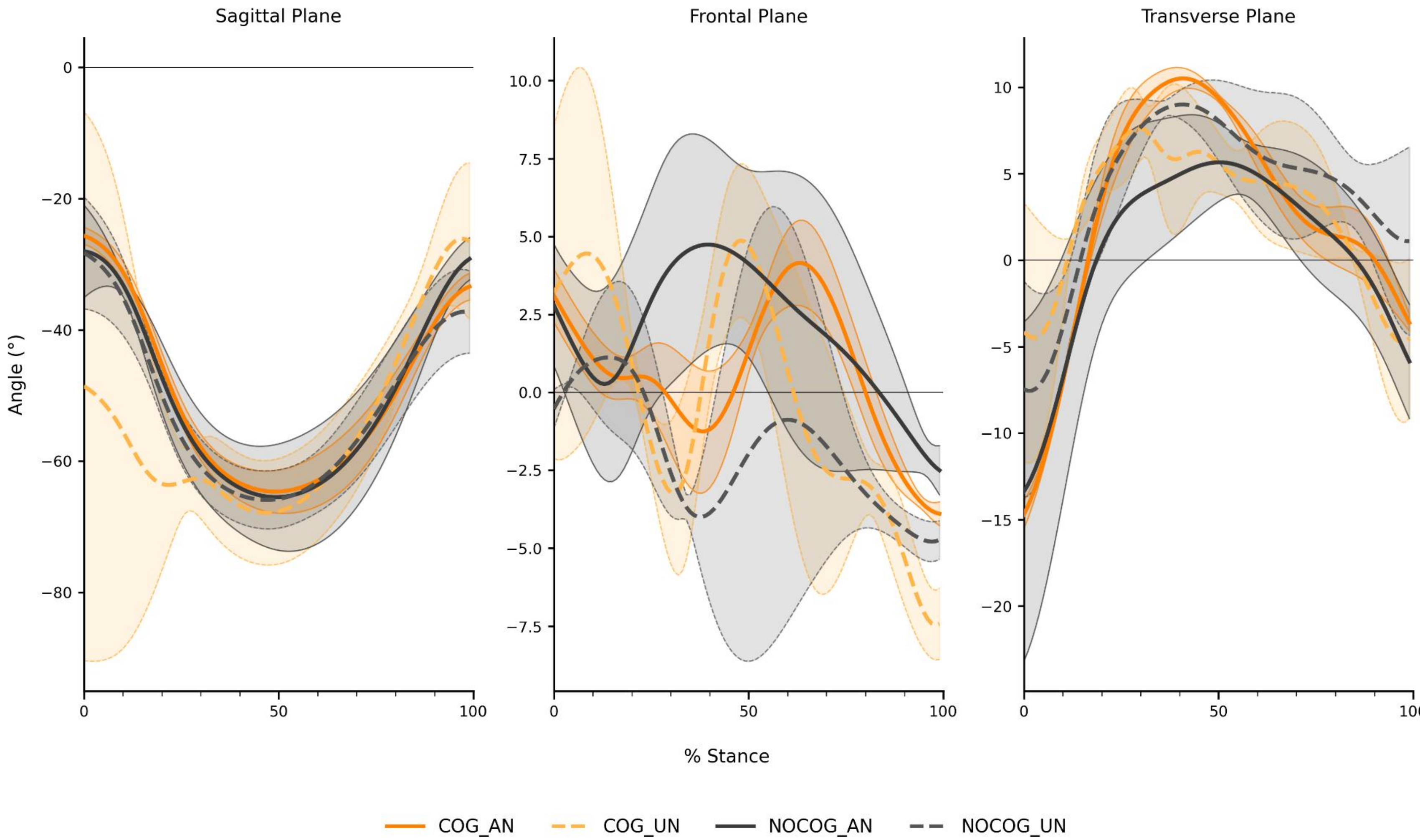

Figure 17. Subject 10 three-dimensional knee joint angles from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee angle for each condition.

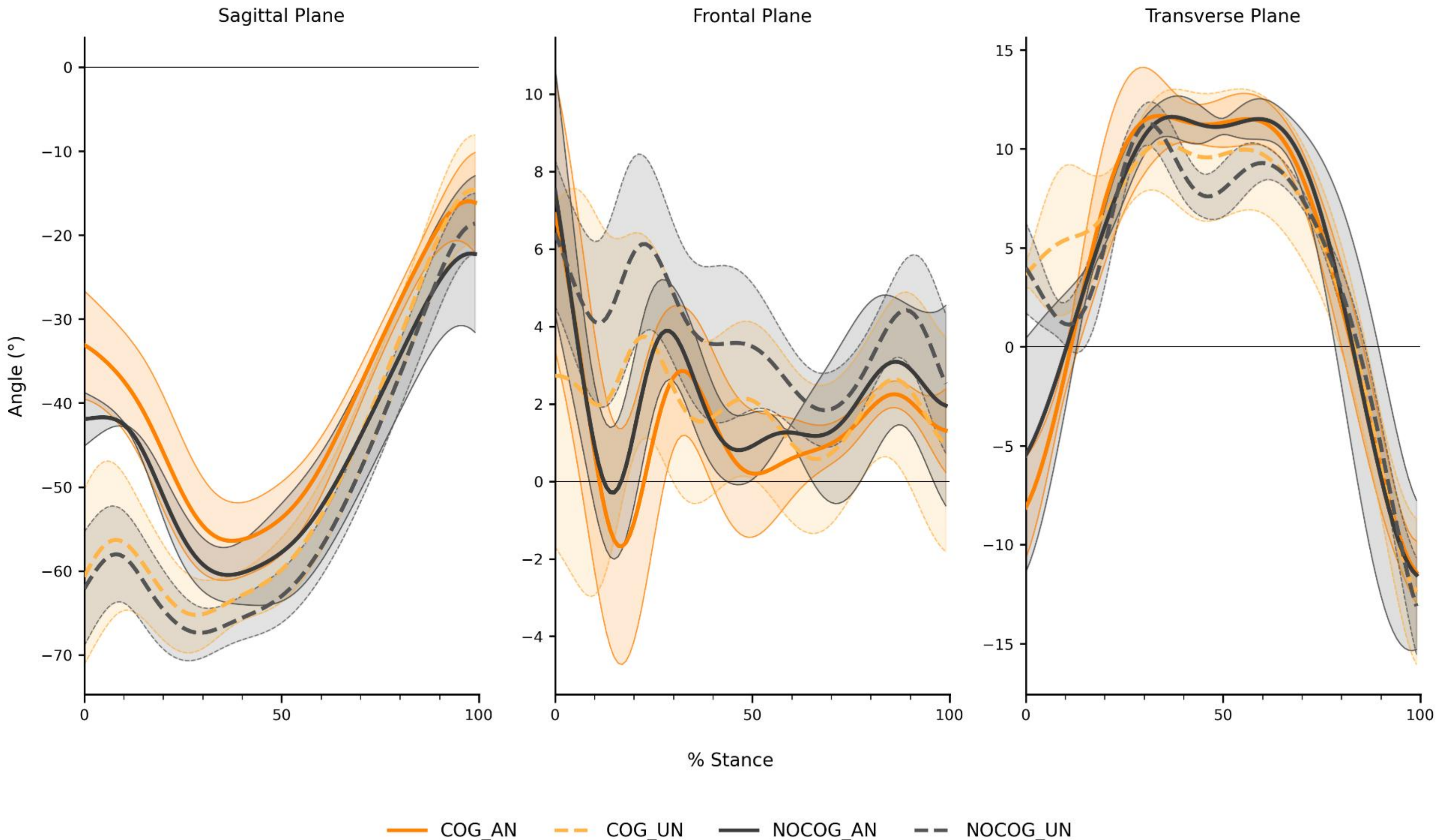

Figure 18. Subject 11 three-dimensional knee joint angles from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee angle for each condition.

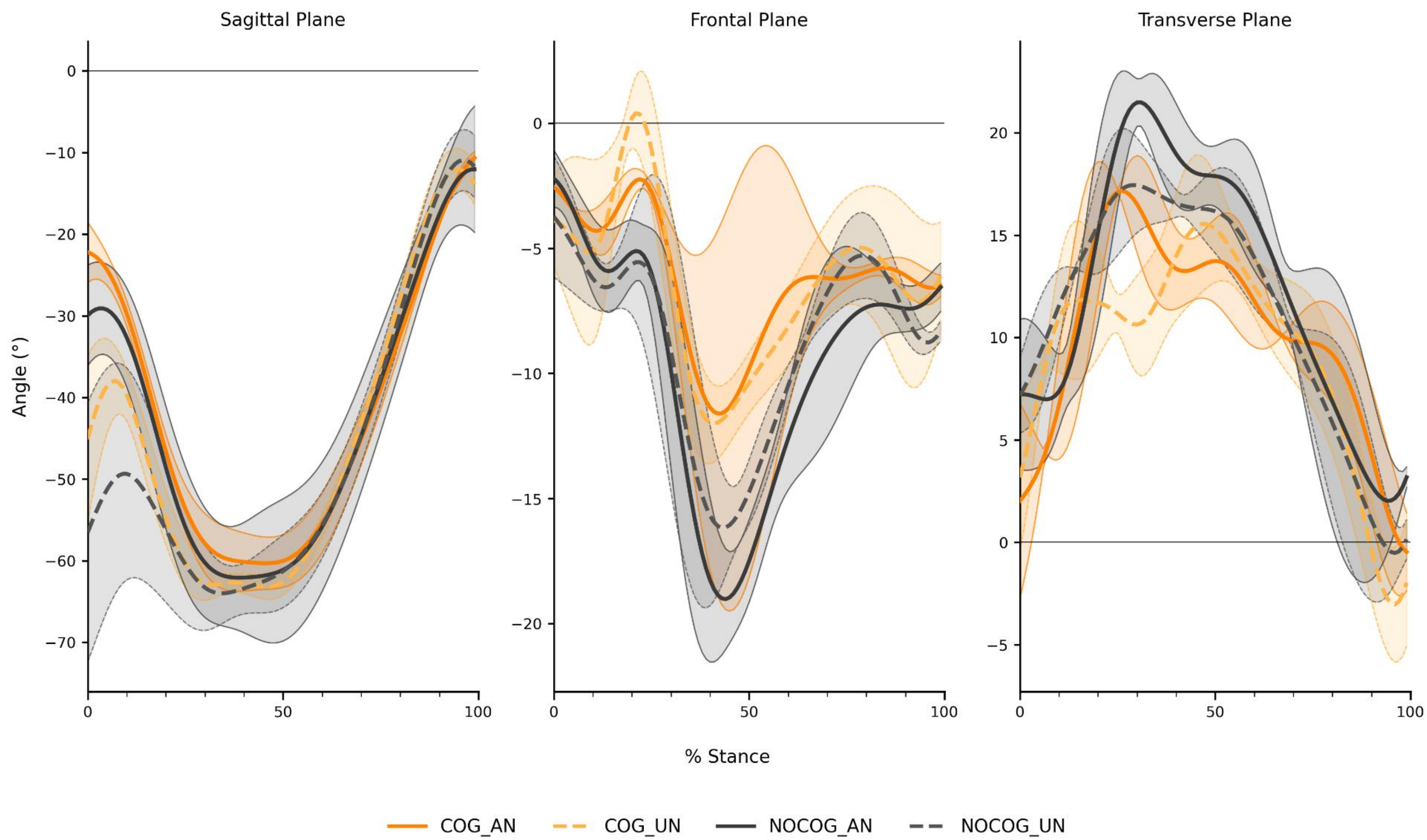

Figure 19. Subject 12 three-dimensional knee joint angles from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee angle for each condition.

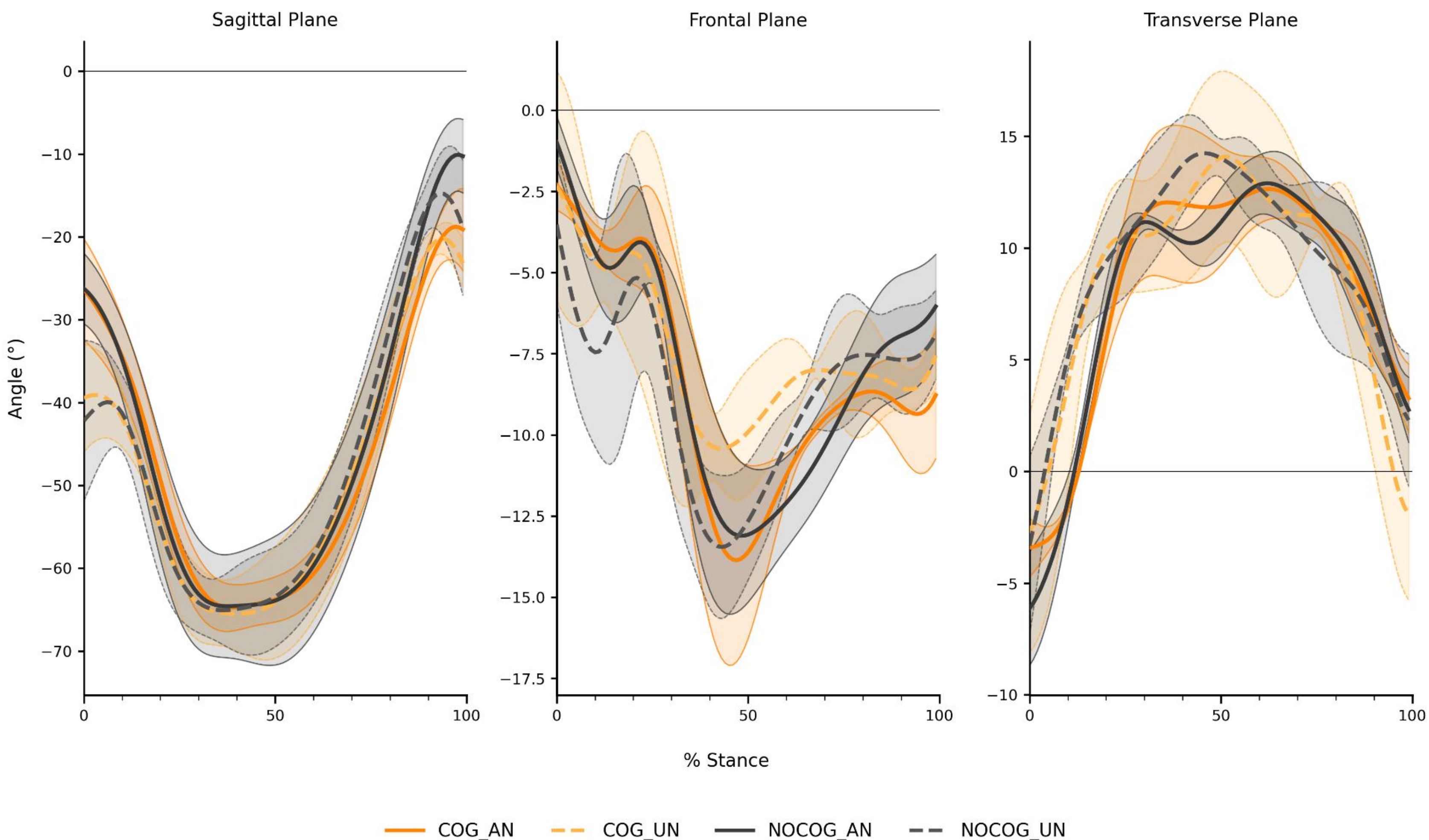

Figure 20. Subject 13 three-dimensional knee joint angles from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee angle for each condition.

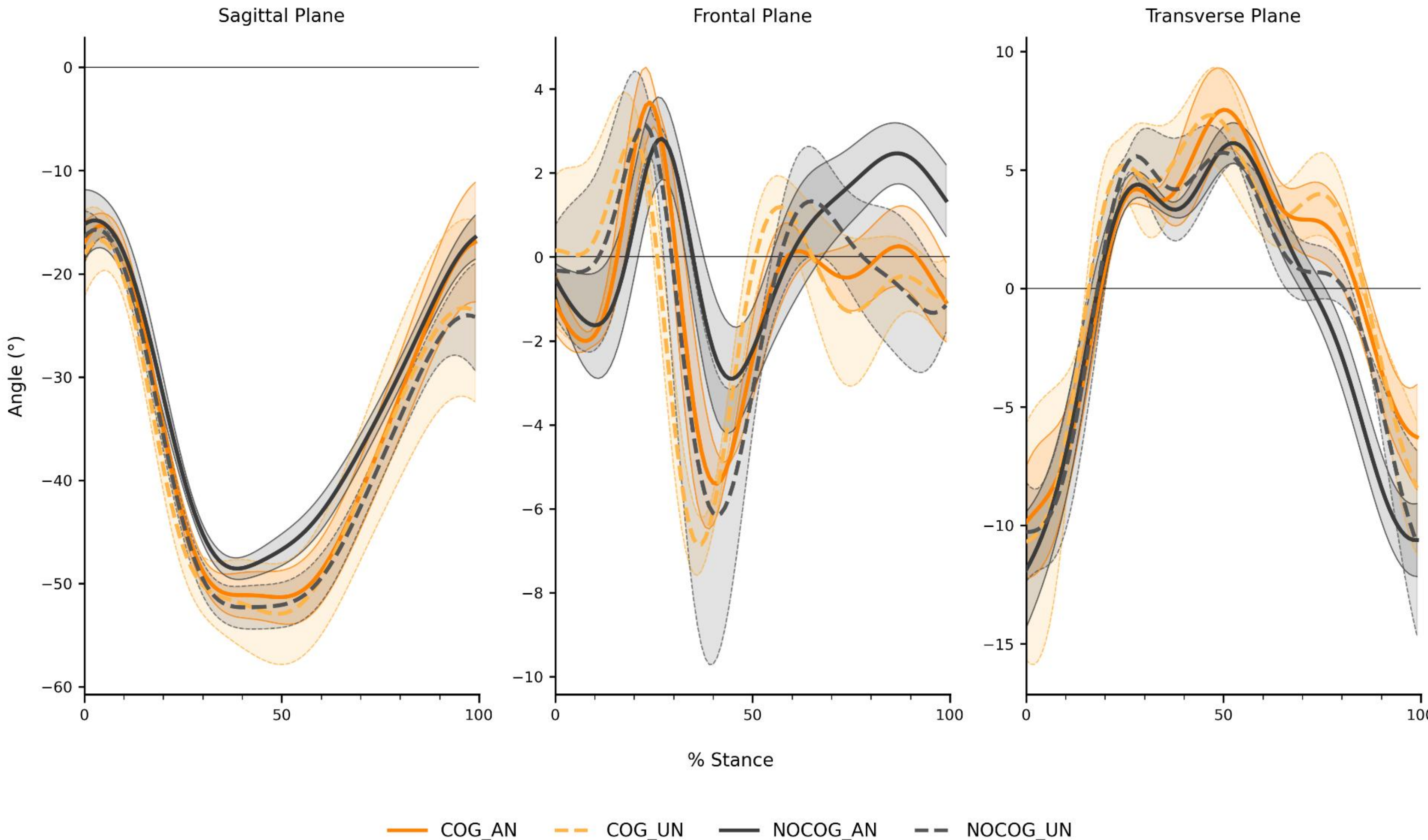

Figure 21. Subject 14 three-dimensional knee joint angles from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee angle for each condition.

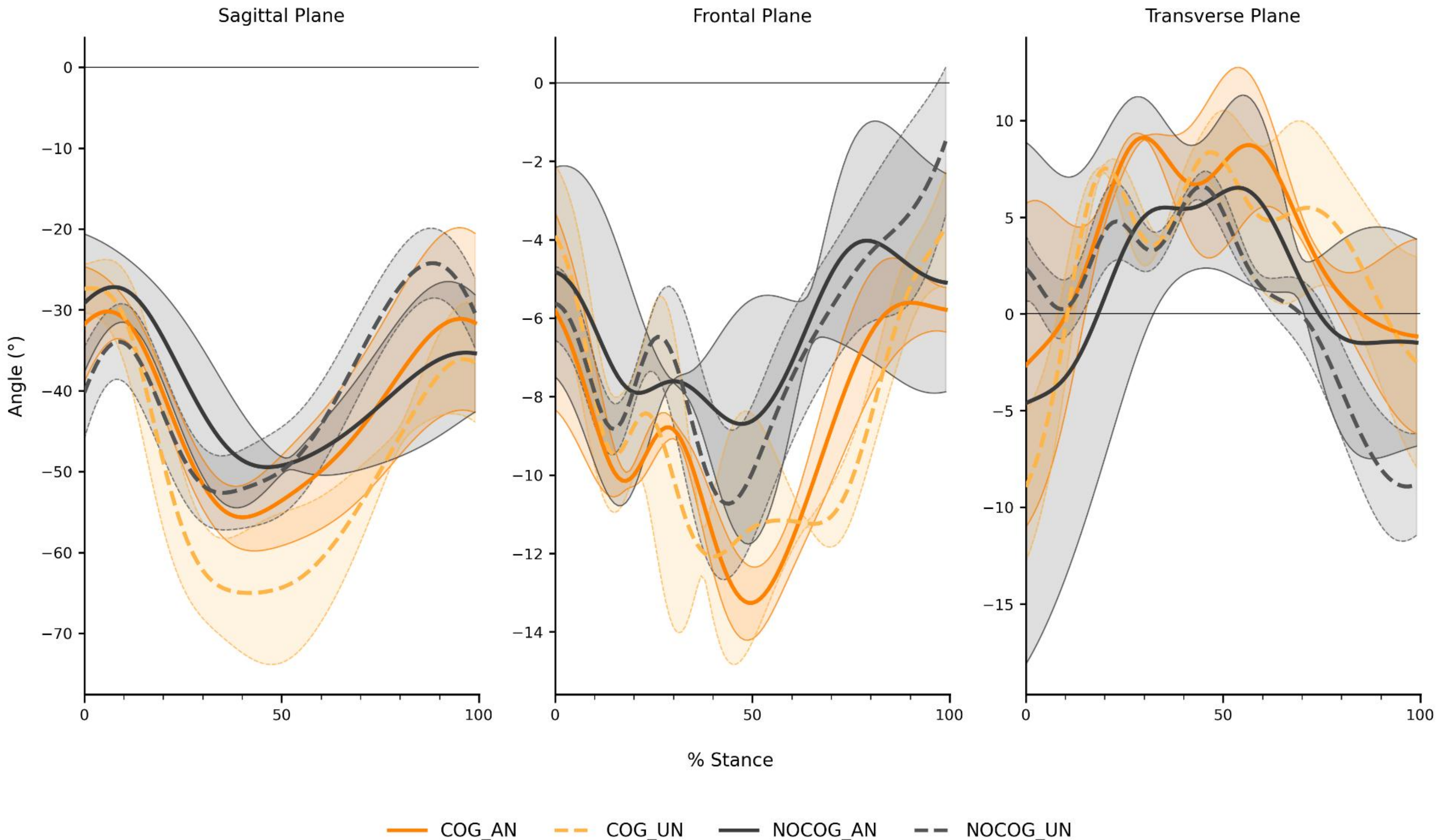

Figure 22. Subject 15 three-dimensional knee joint angles from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee angle for each condition.

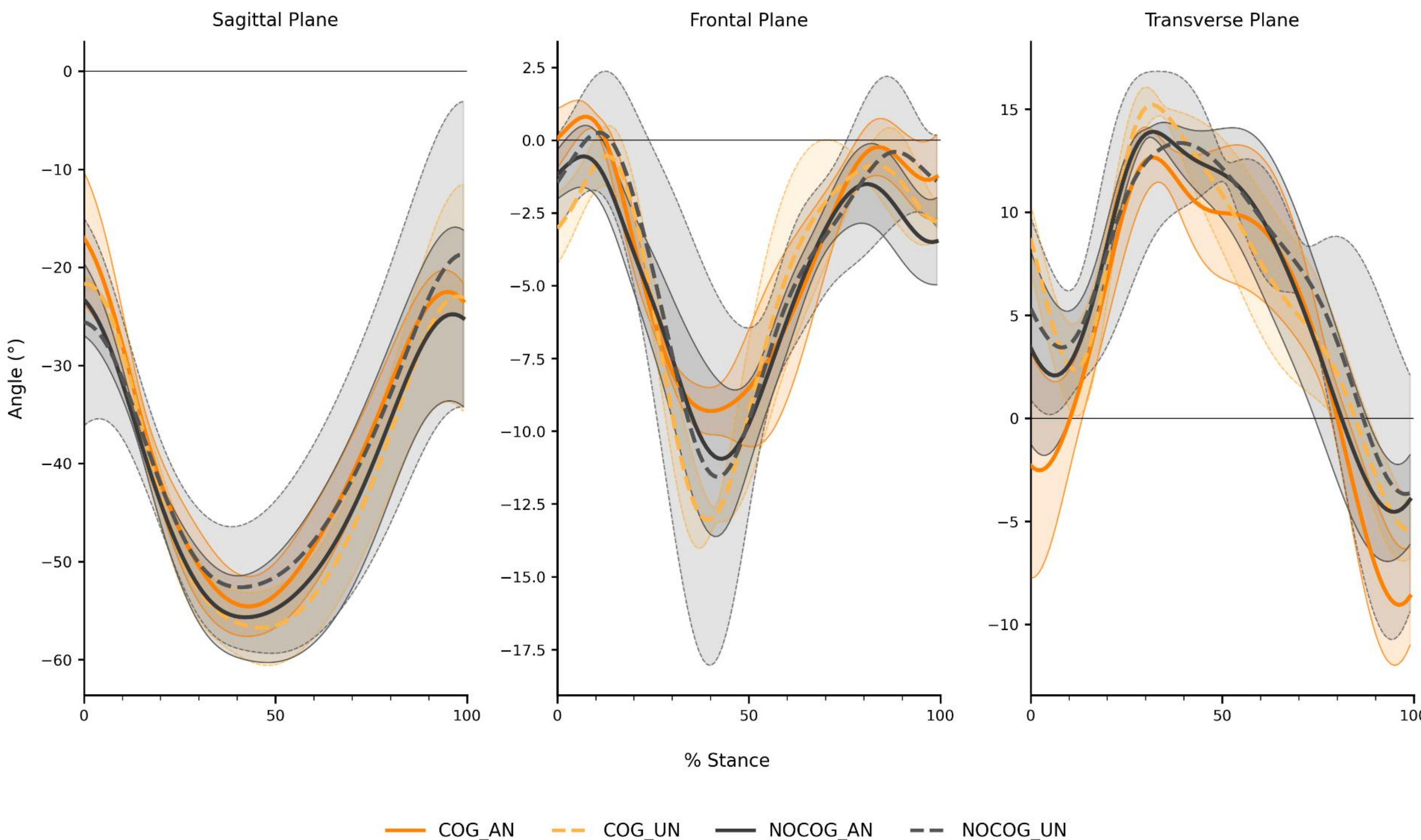

Figure 23. Subject 16 three-dimensional knee joint angles from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee angle for each condition.

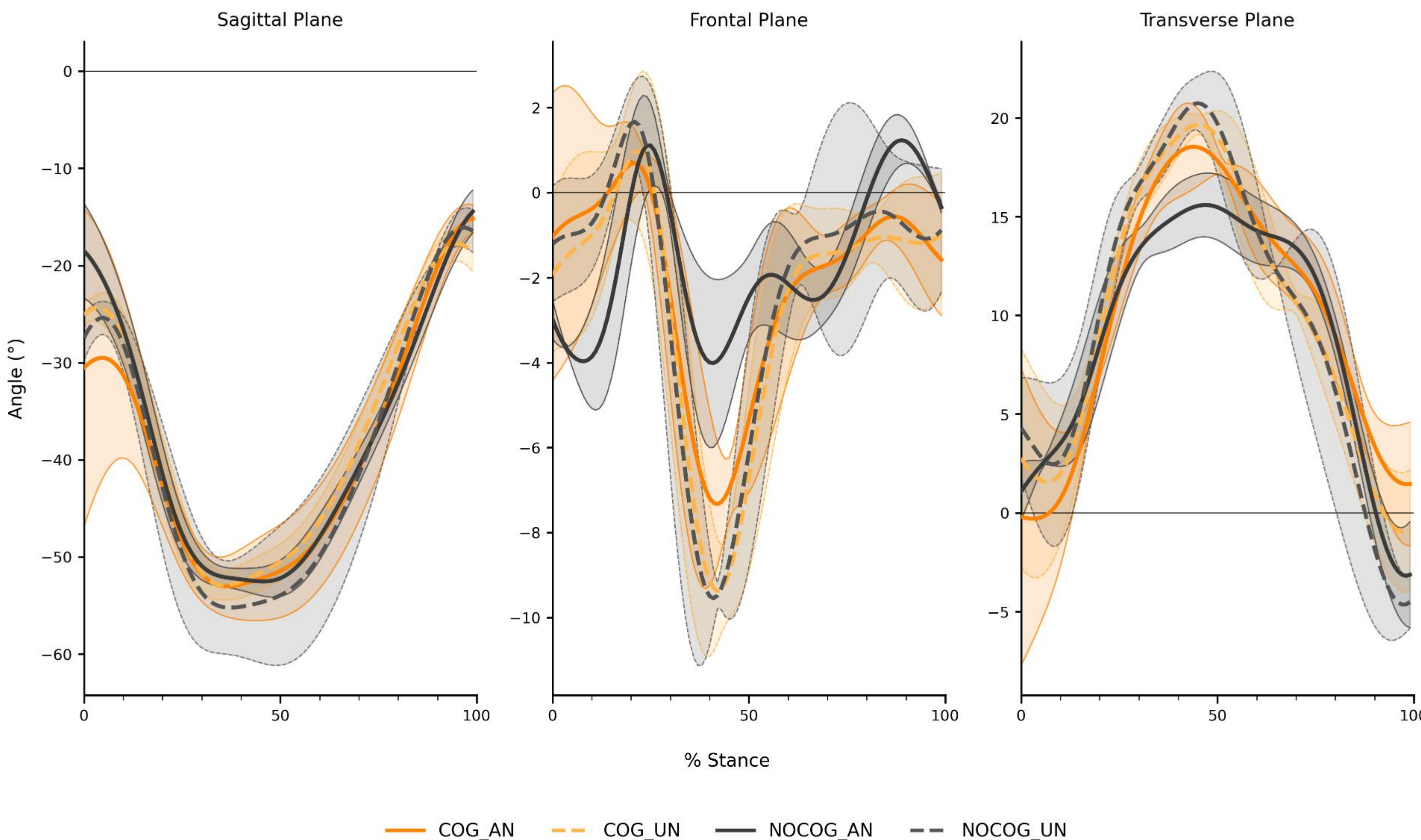

Figure 24. Subject 17 three-dimensional knee joint angles from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee angle for each condition.

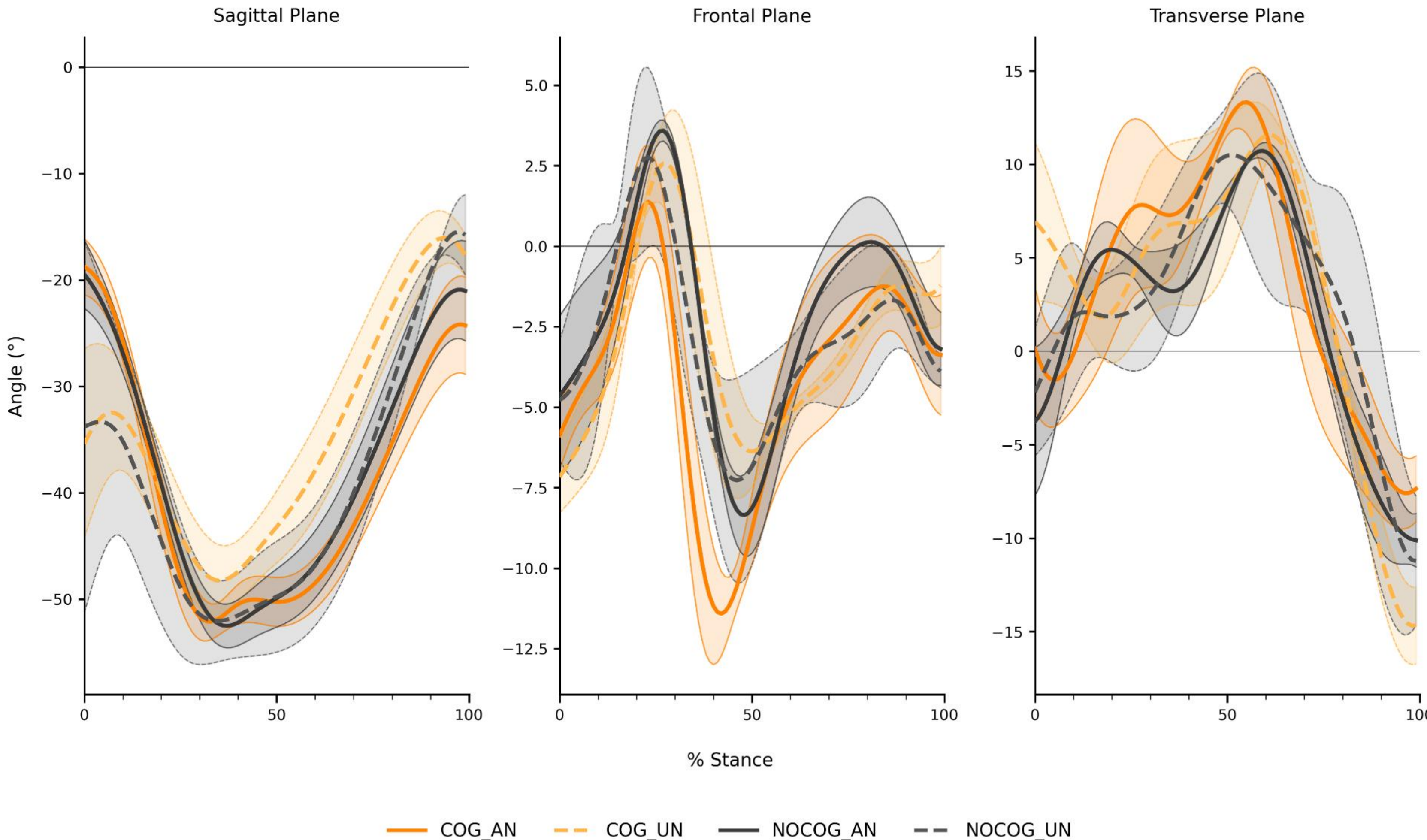

Figure 25. Subject 18 three-dimensional knee joint angles from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee angle for each condition.

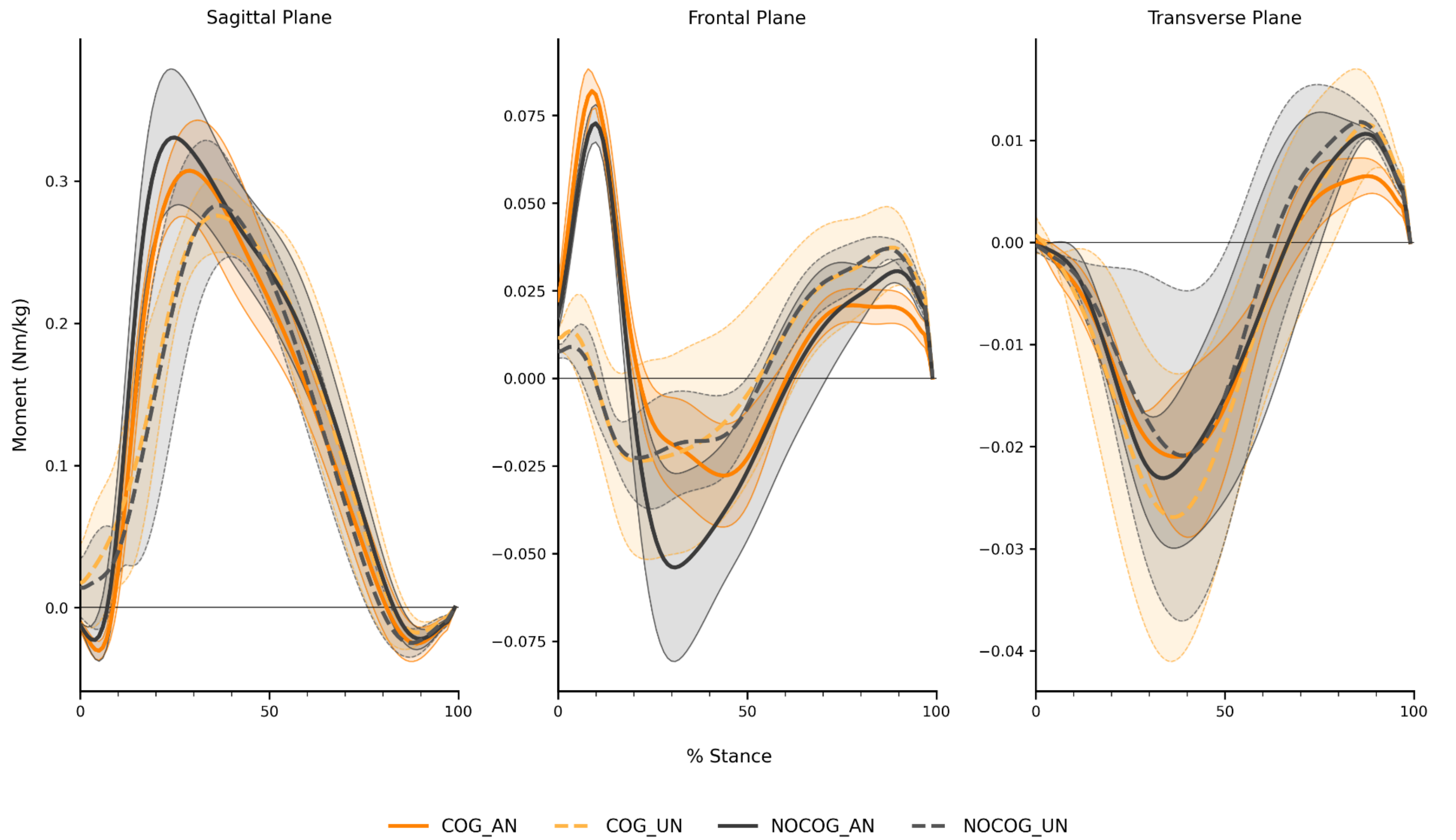

Figure 26. Subject 01 three-dimensional knee joint moments from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee moment for each condition.

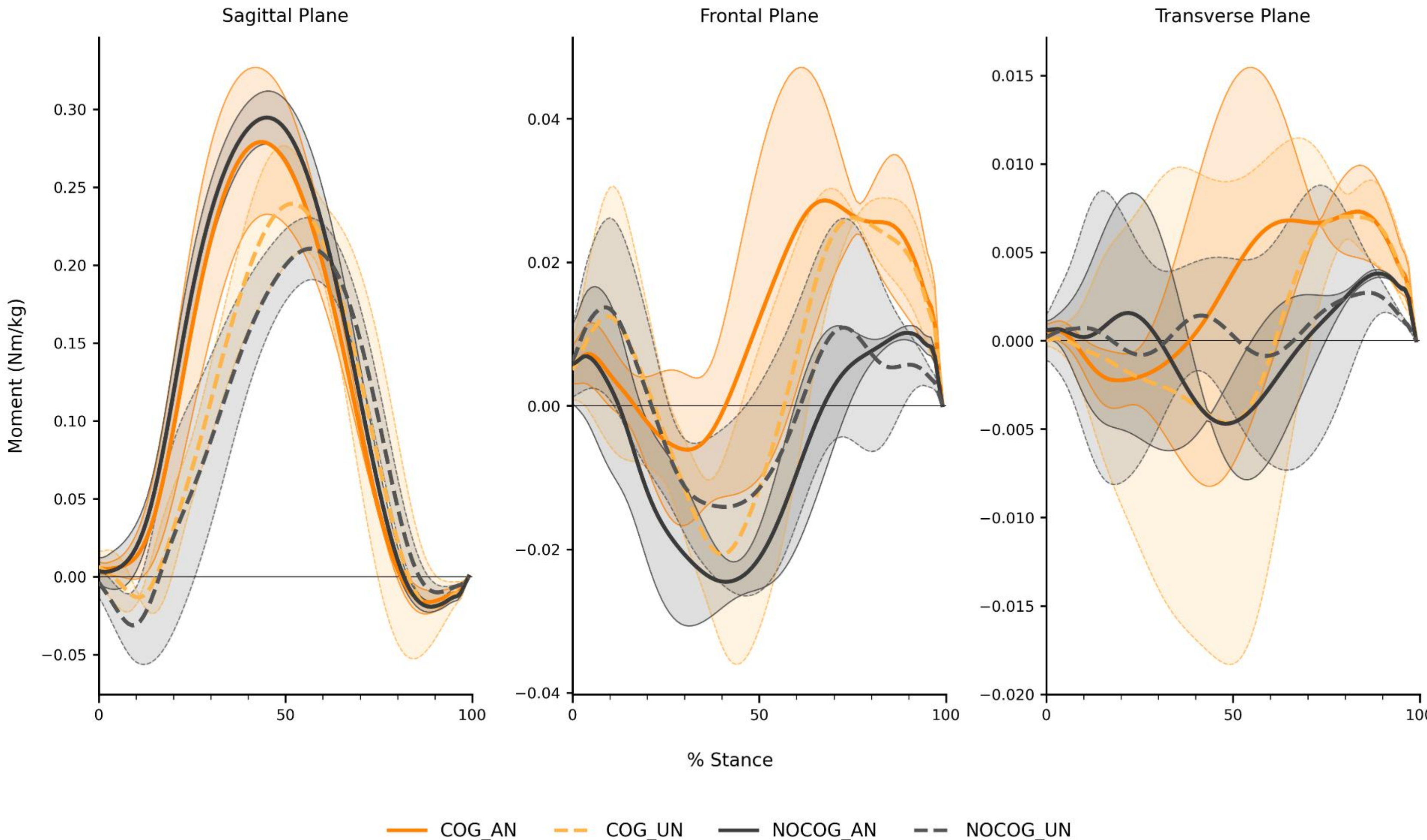

Figure 27. Subject 03 three-dimensional knee joint moments from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee moment for each condition.

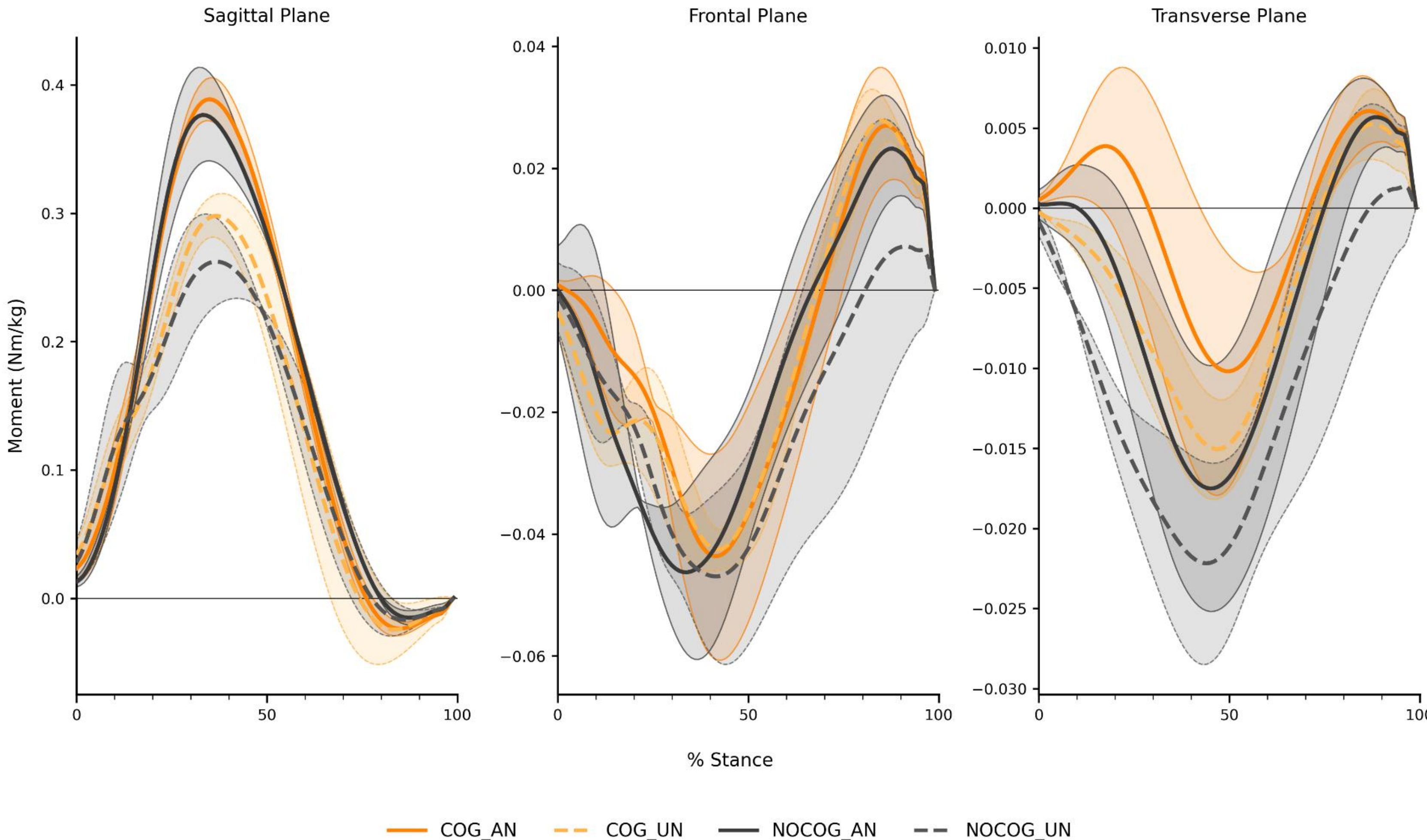

Figure 28. Subject 04 three-dimensional knee joint moments from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee moment for each condition.

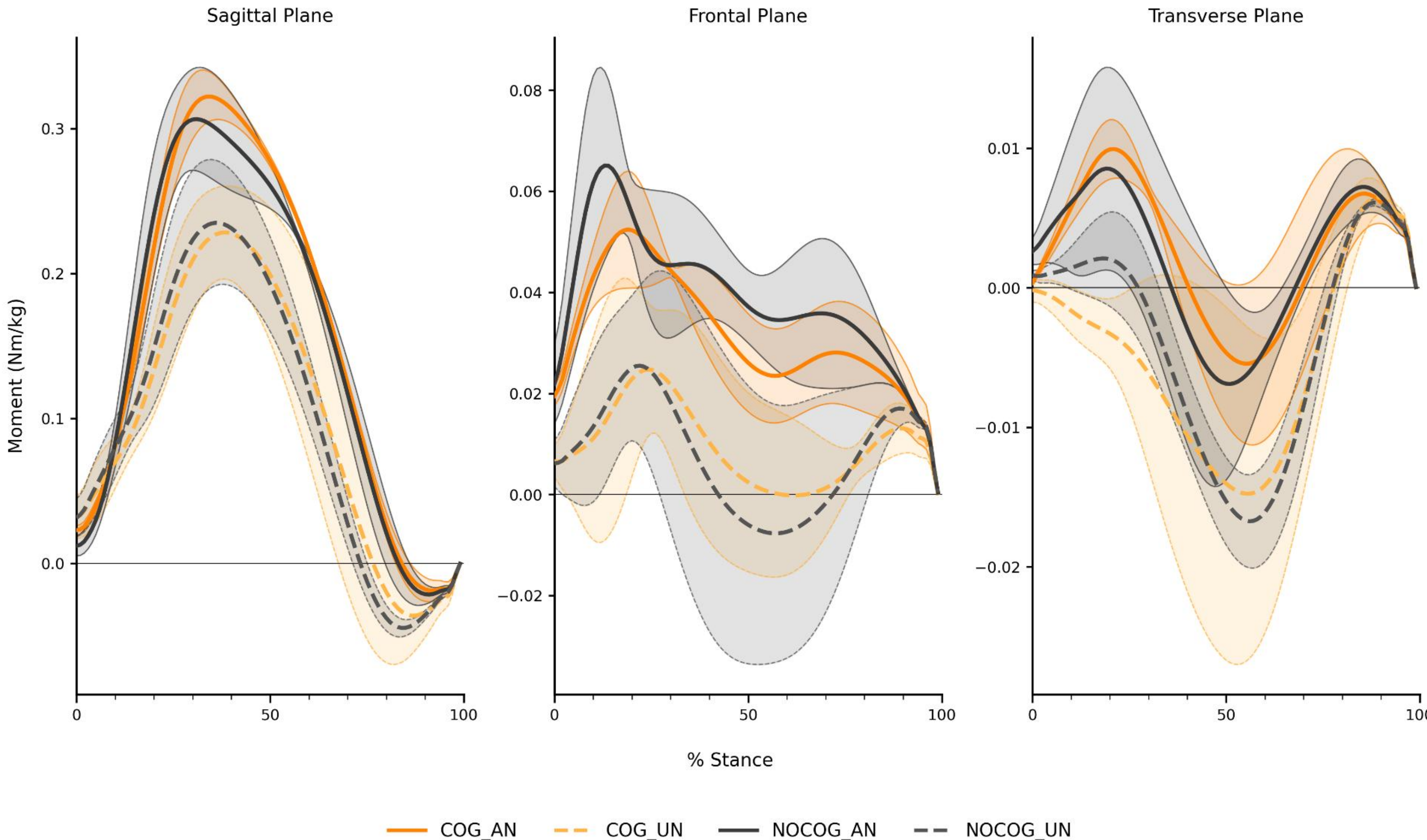

Figure 29. Subject 05 three-dimensional knee joint moments from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee moment for each condition.

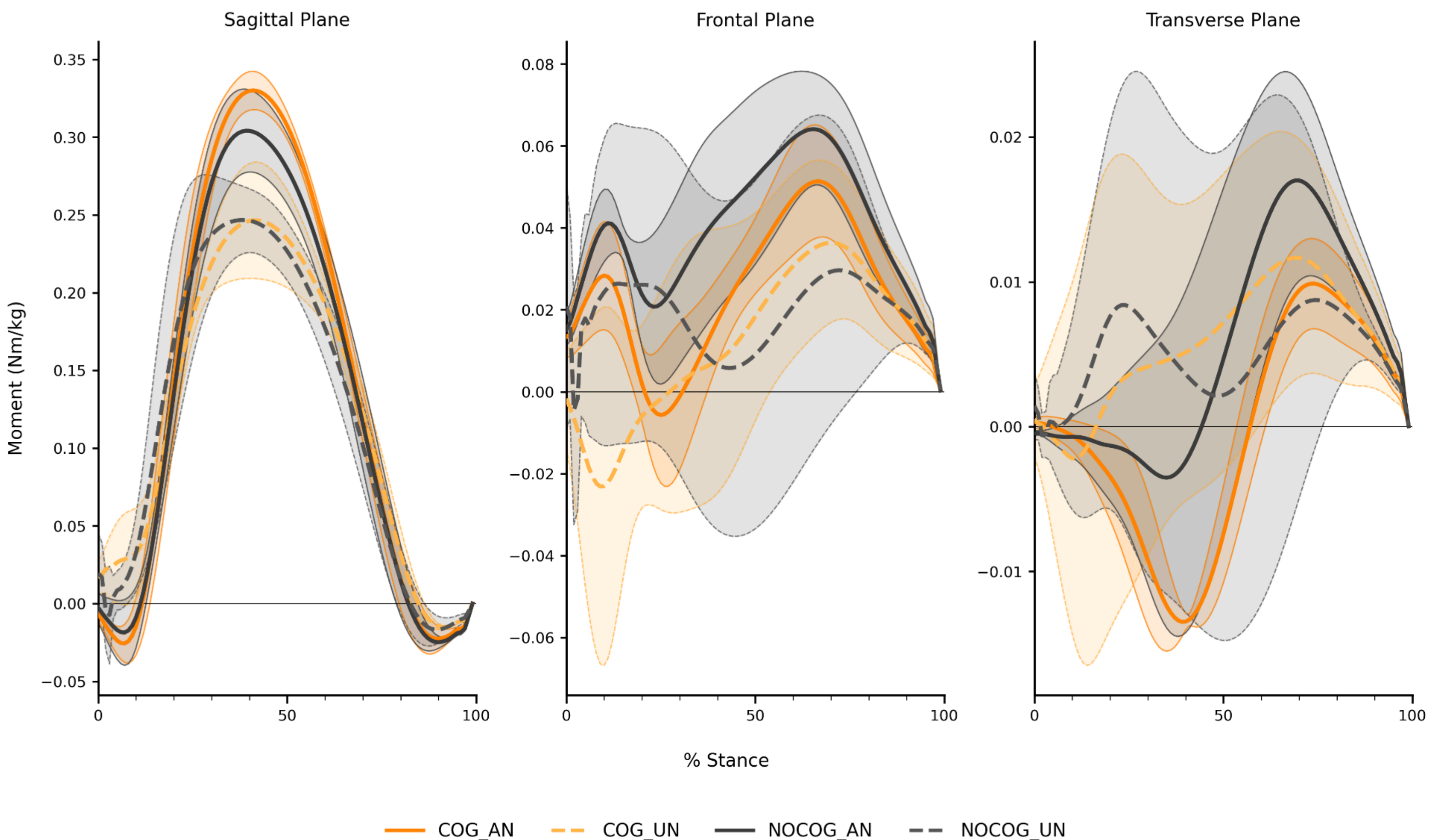

Figure 30. Subject 06 three-dimensional knee joint moments from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee moment for each condition.

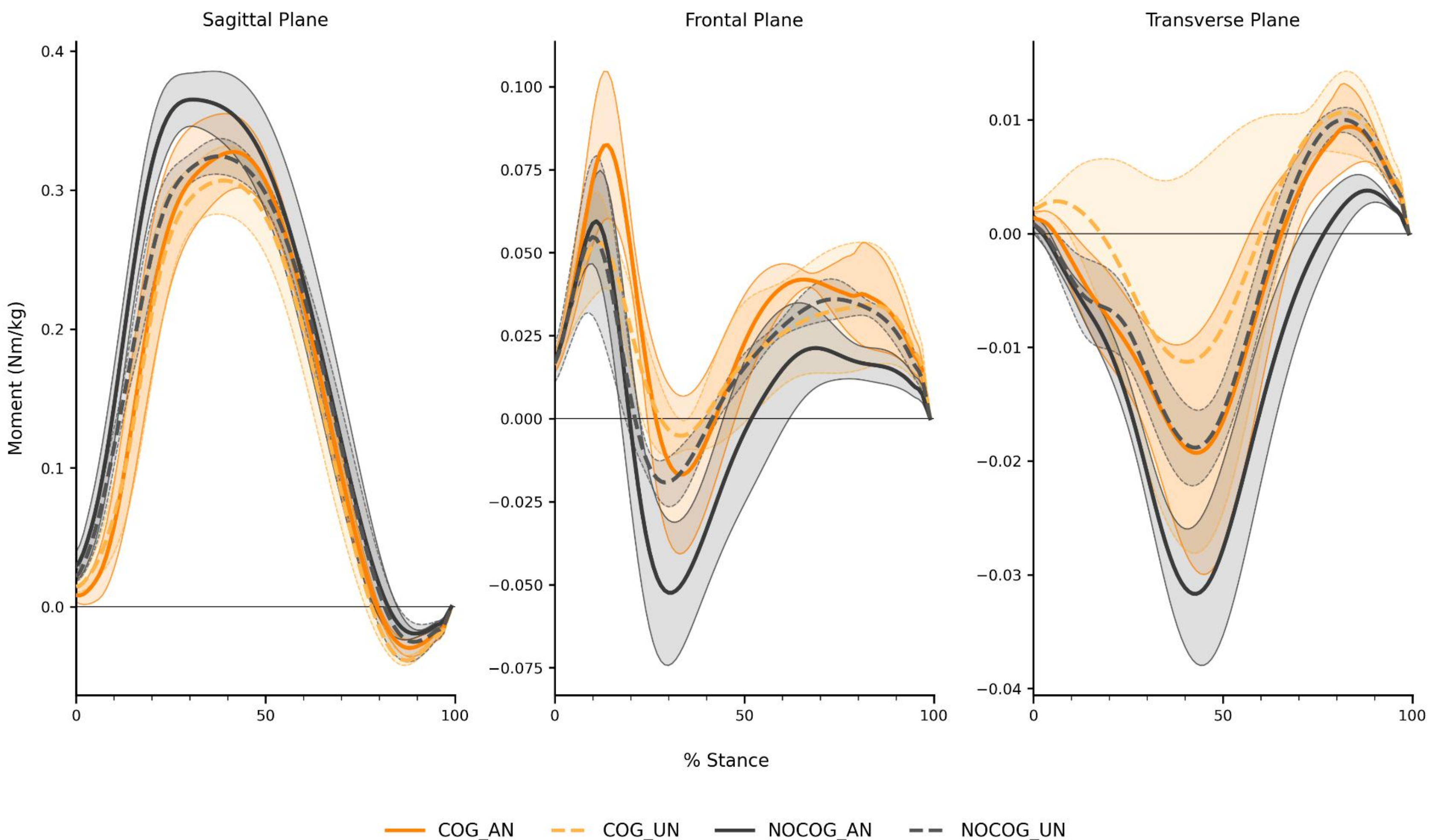

Figure 31. Subject 07 three-dimensional knee joint moments from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee moment for each condition.

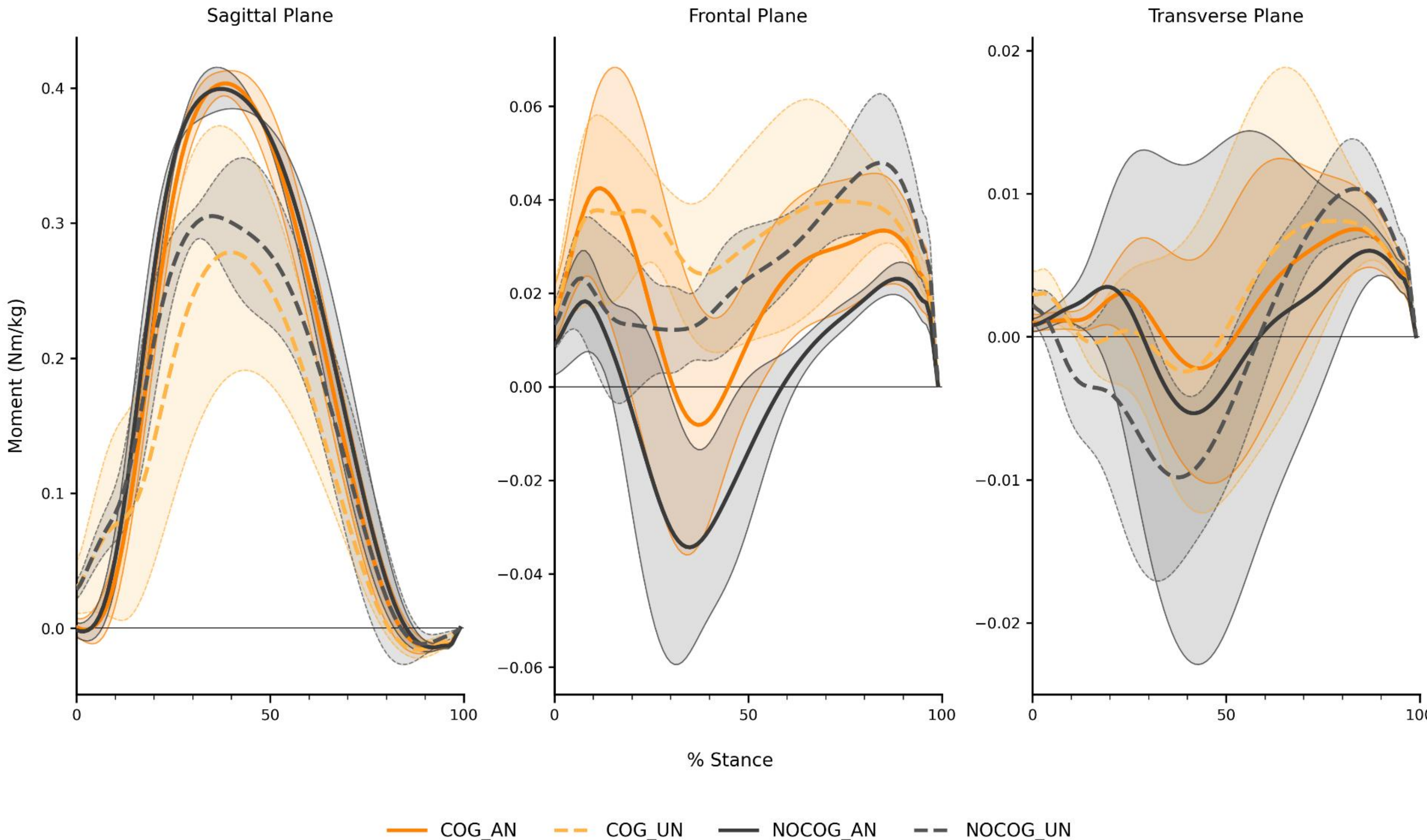

Figure 32. Subject 08 three-dimensional knee joint moments from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee moment for each condition.

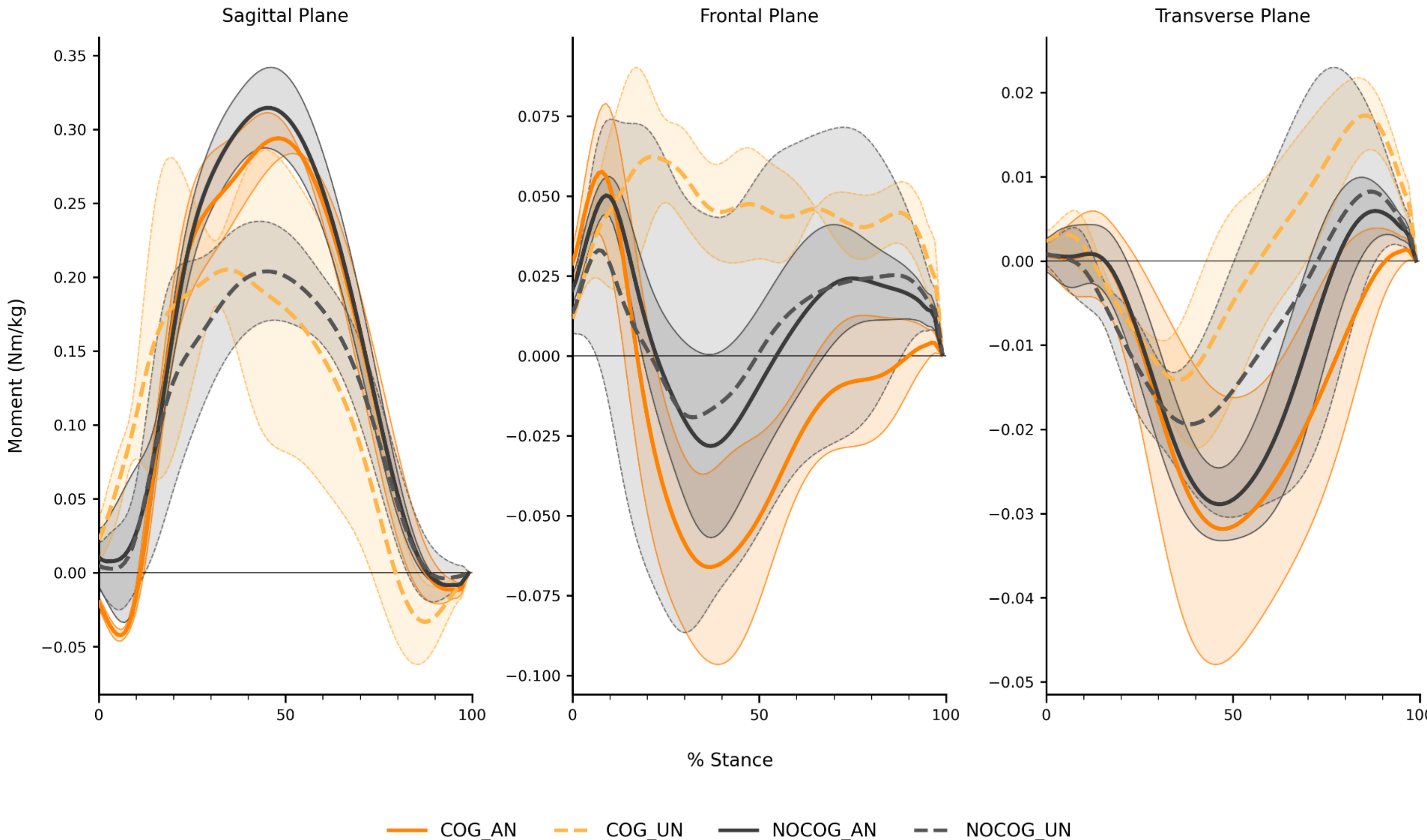

Figure 33. Subject 09 three-dimensional knee joint moments from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee moment for each condition.

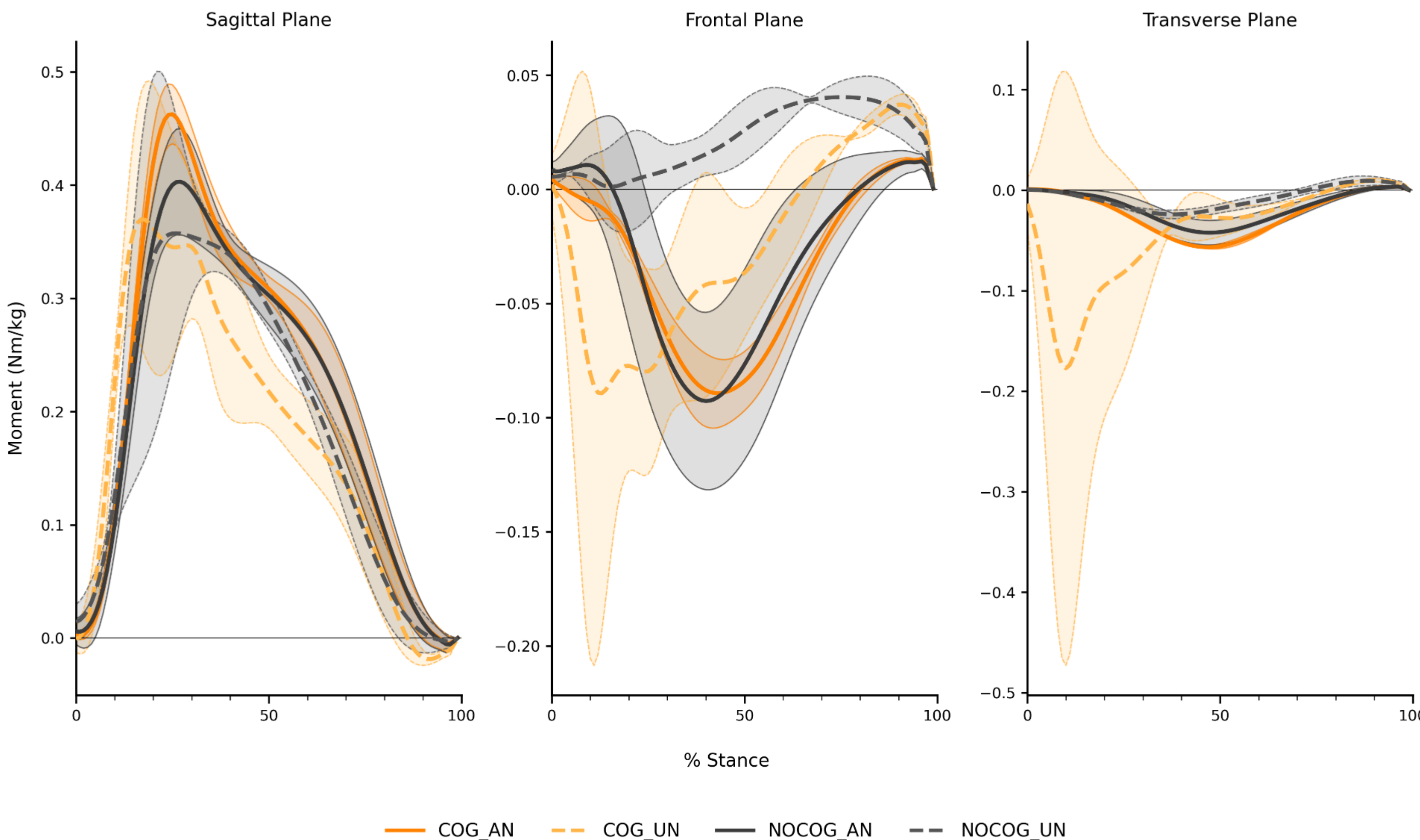

Figure 34. Subject 10 three-dimensional knee joint moments from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee moment for each condition.

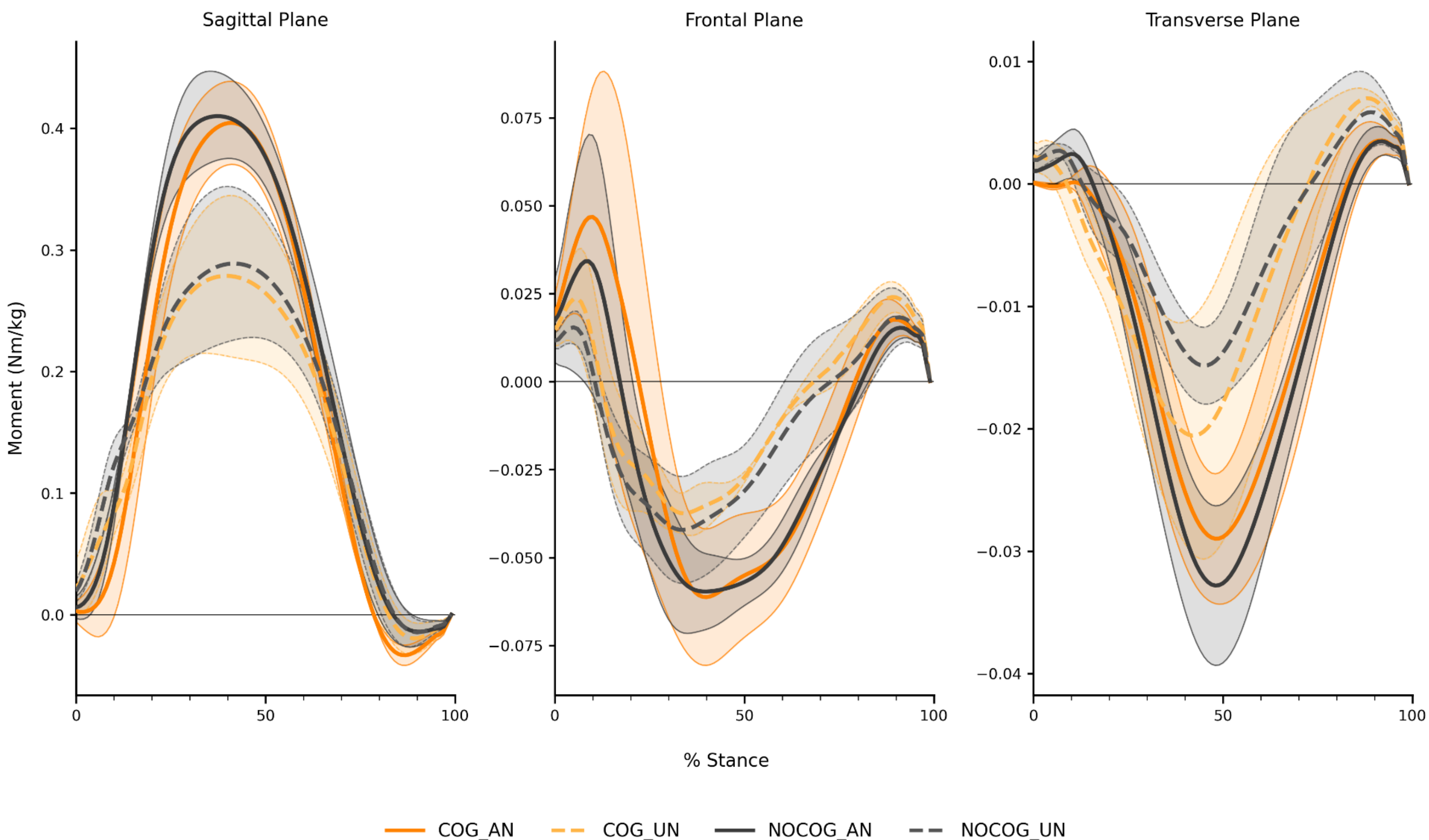

Figure 35. Subject 11 three-dimensional knee joint moments from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee moment for each condition.

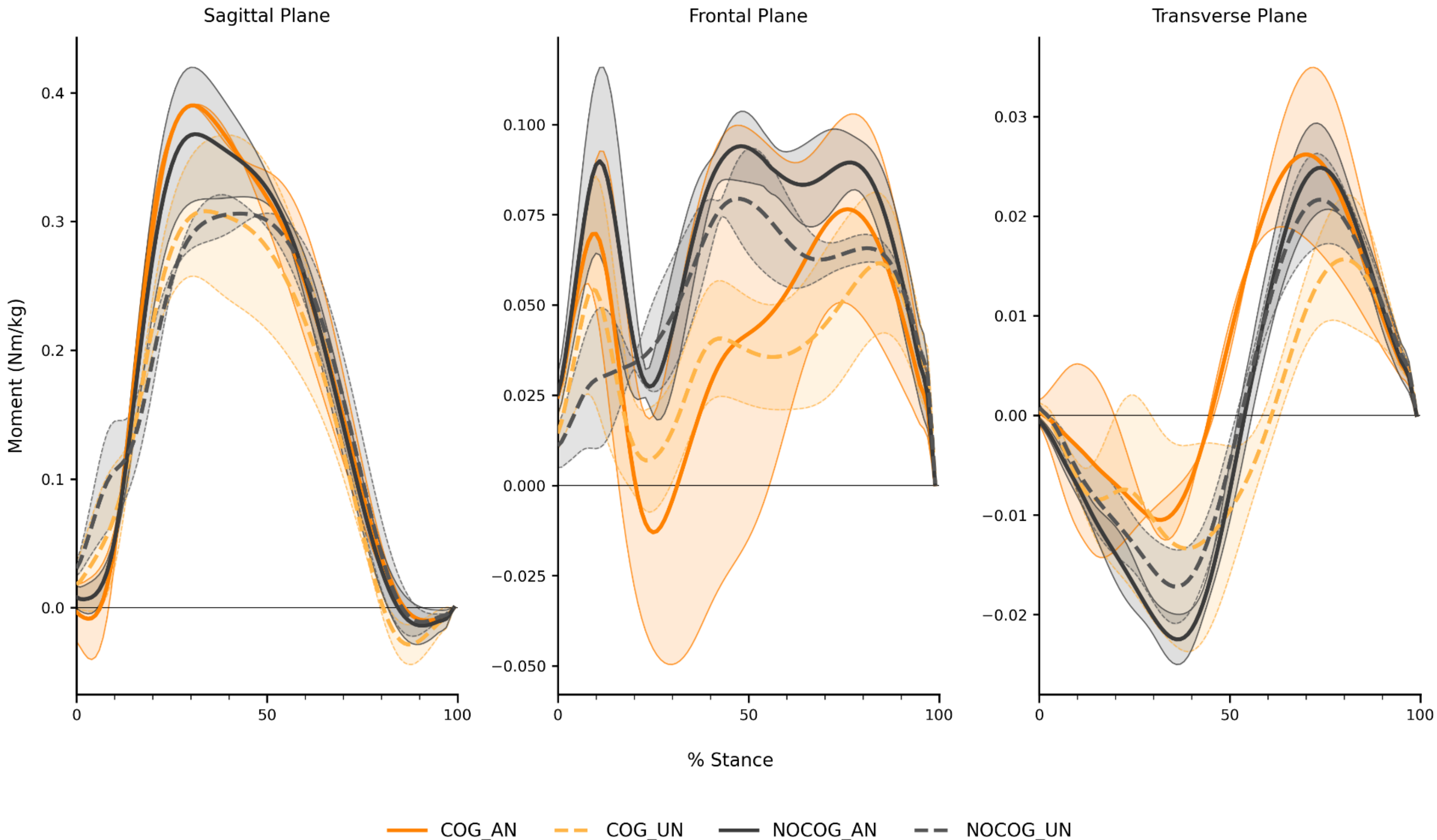

Figure 36. Subject 12 three-dimensional knee joint moments from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee moment for each condition.

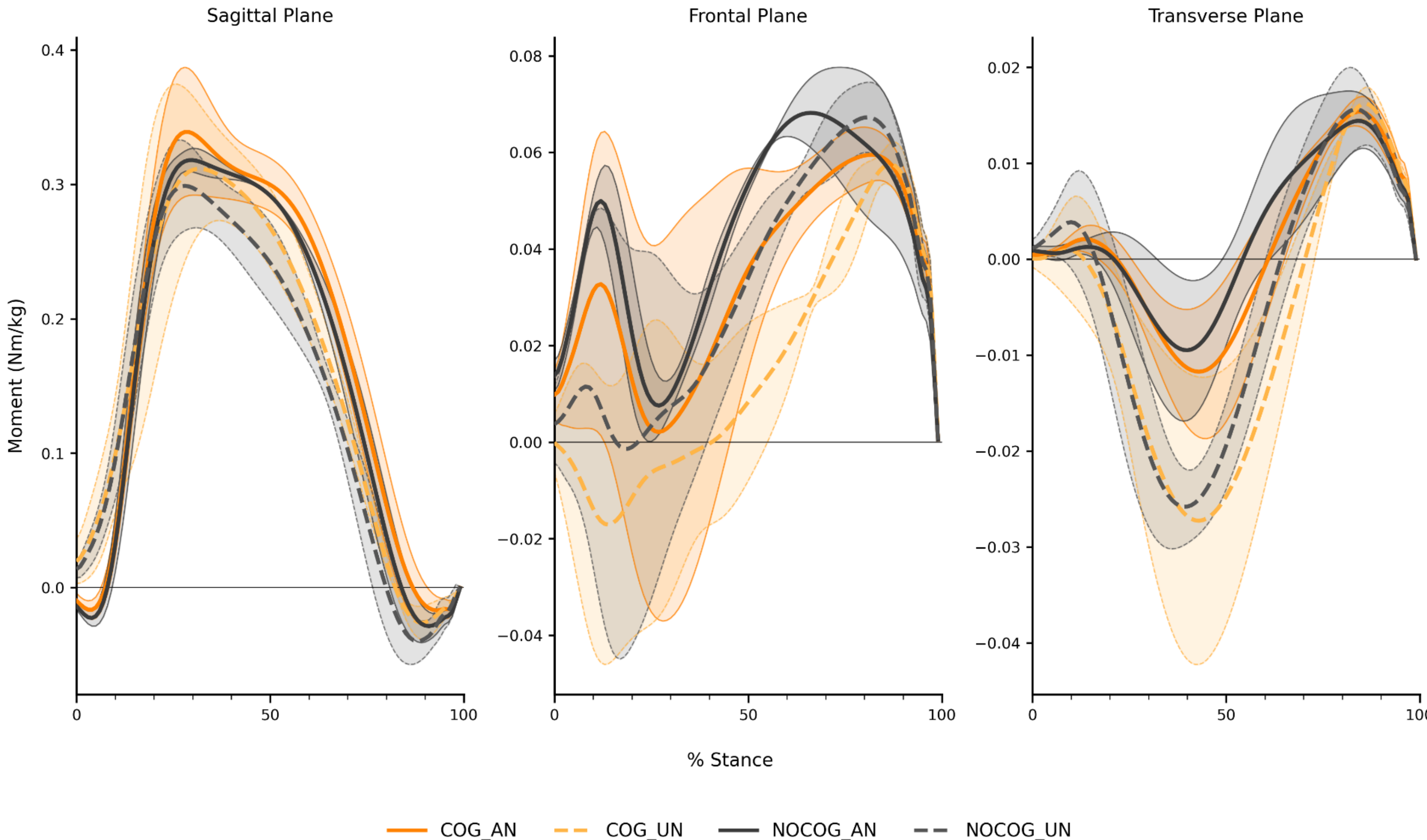

Figure 37. Subject 13 three-dimensional knee joint moments from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee moment for each condition.

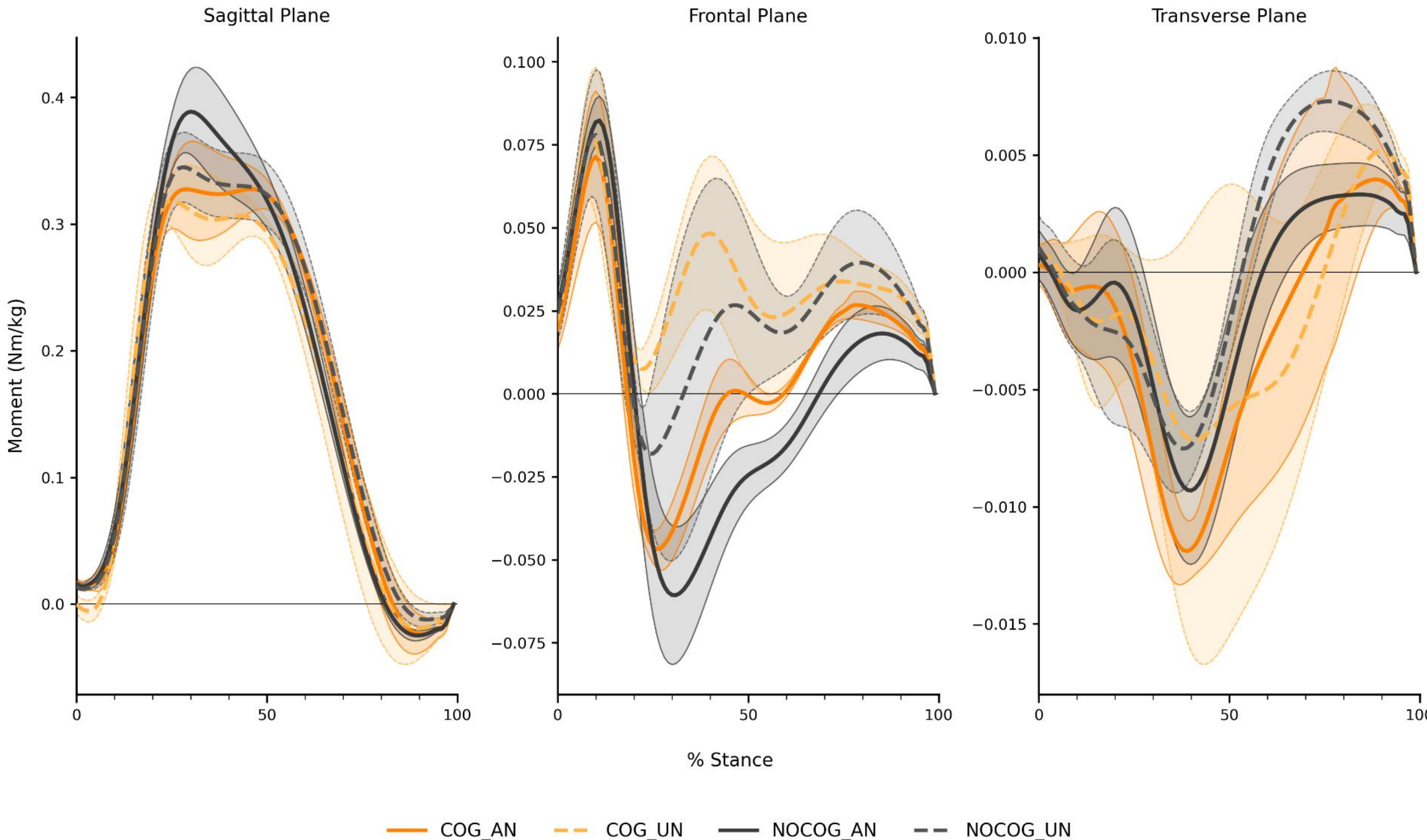

Figure 38. Subject 14 three-dimensional knee joint moments from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee moment for each condition.

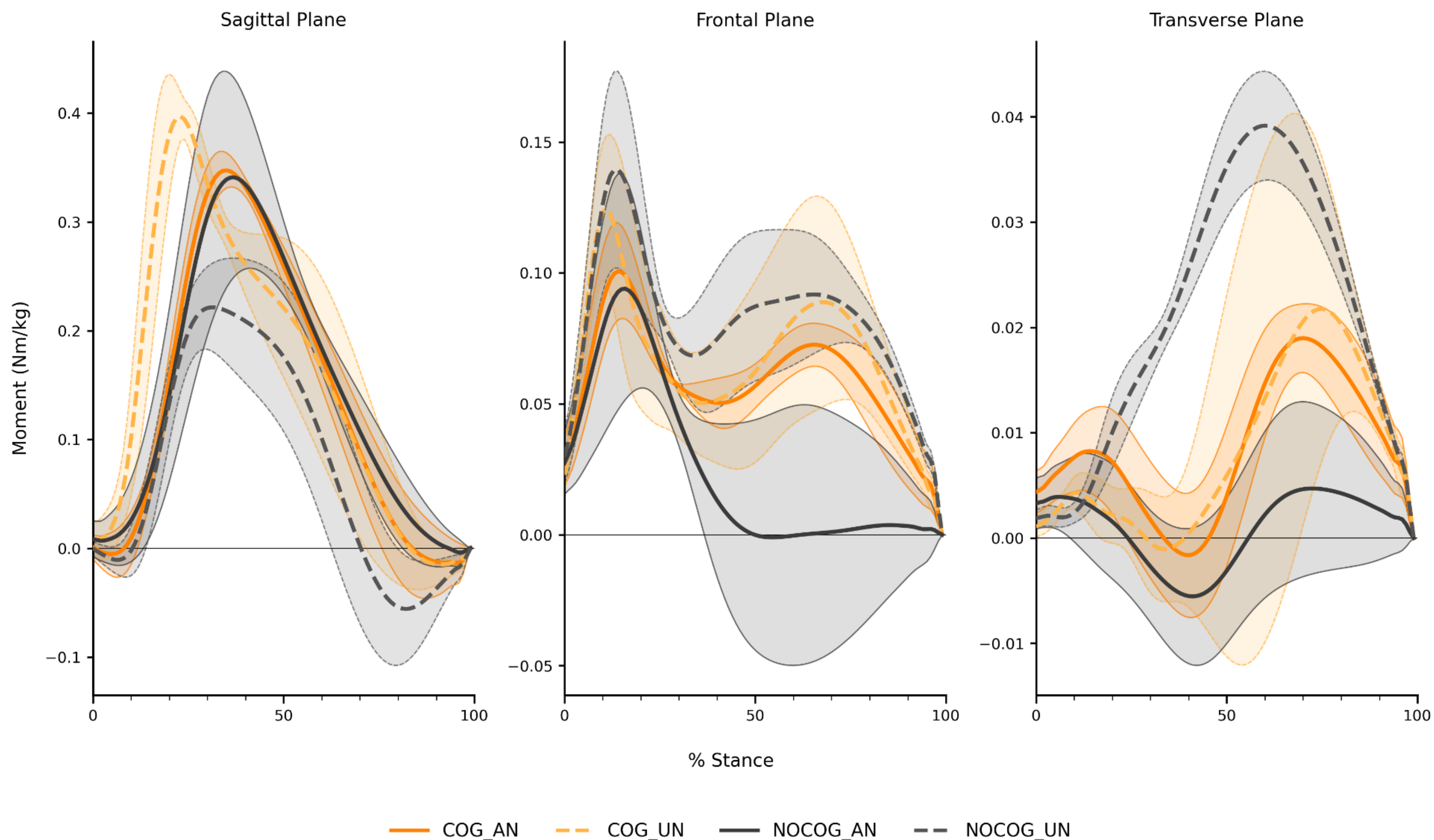

Figure 39. Subject 15 three-dimensional knee joint moments from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee moment for each condition.

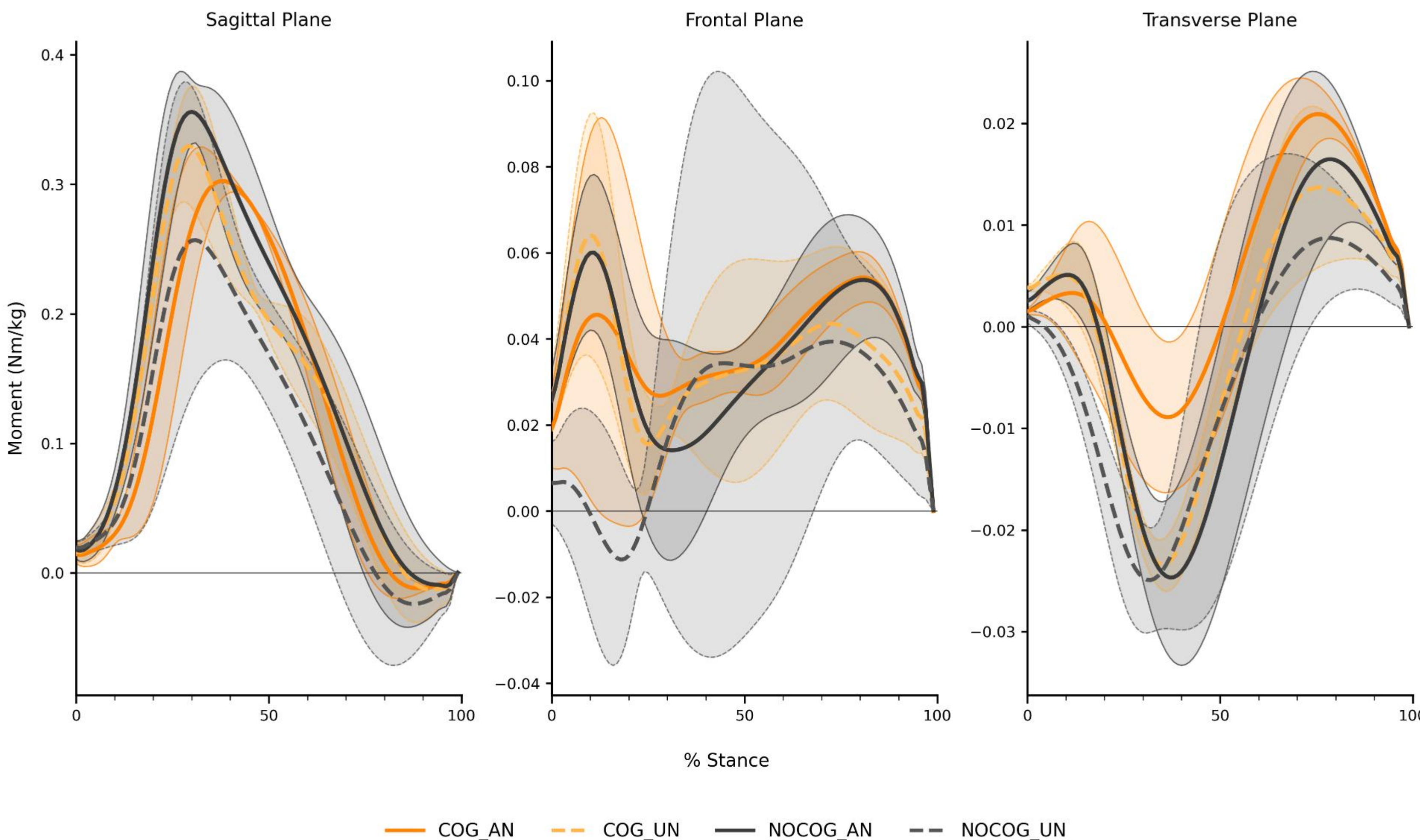

Figure 40. Subject 16 three-dimensional knee joint moments from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee moment for each condition.

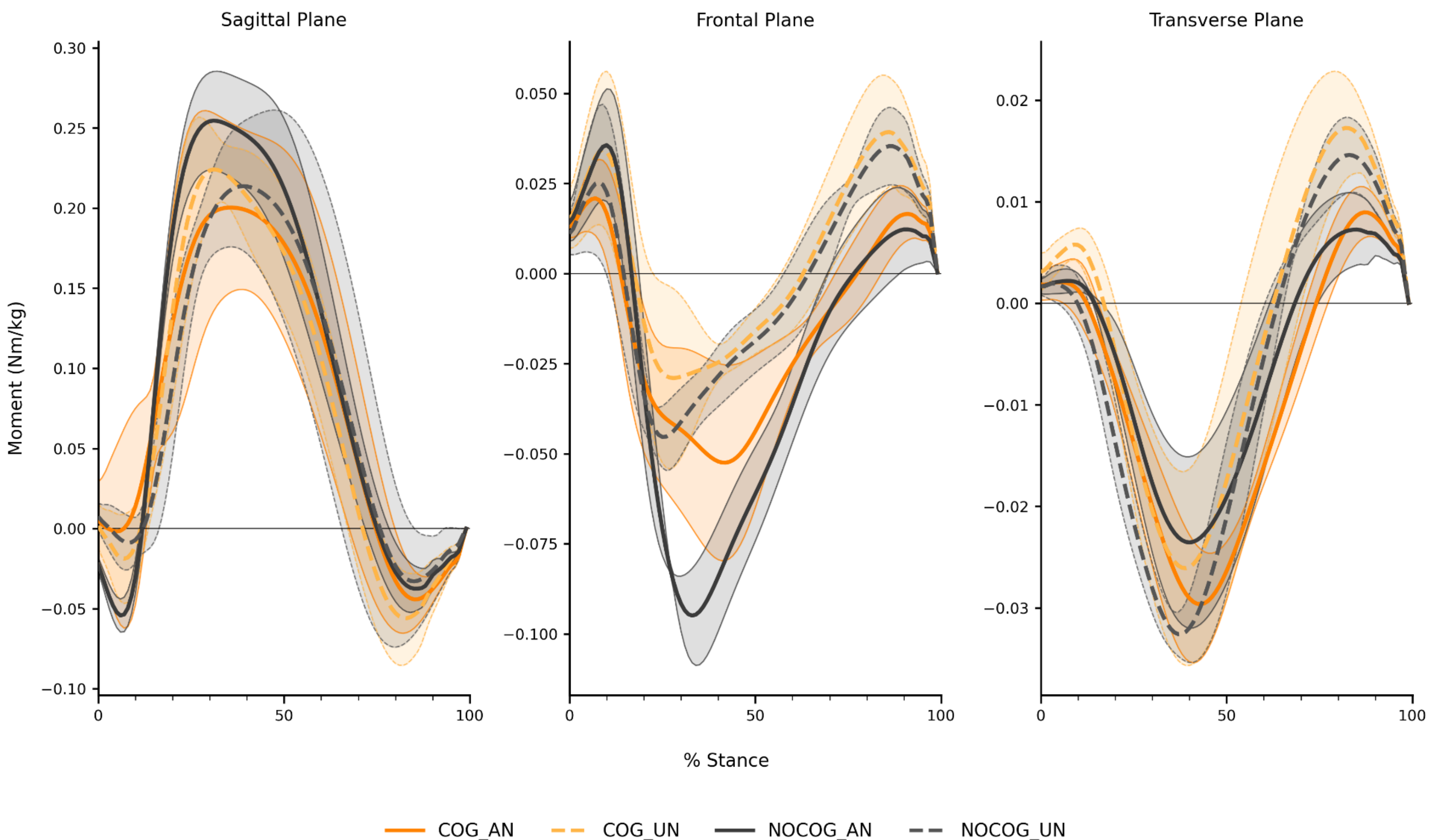

Figure 41. Subject 17 three-dimensional knee joint moments from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee moment for each condition.

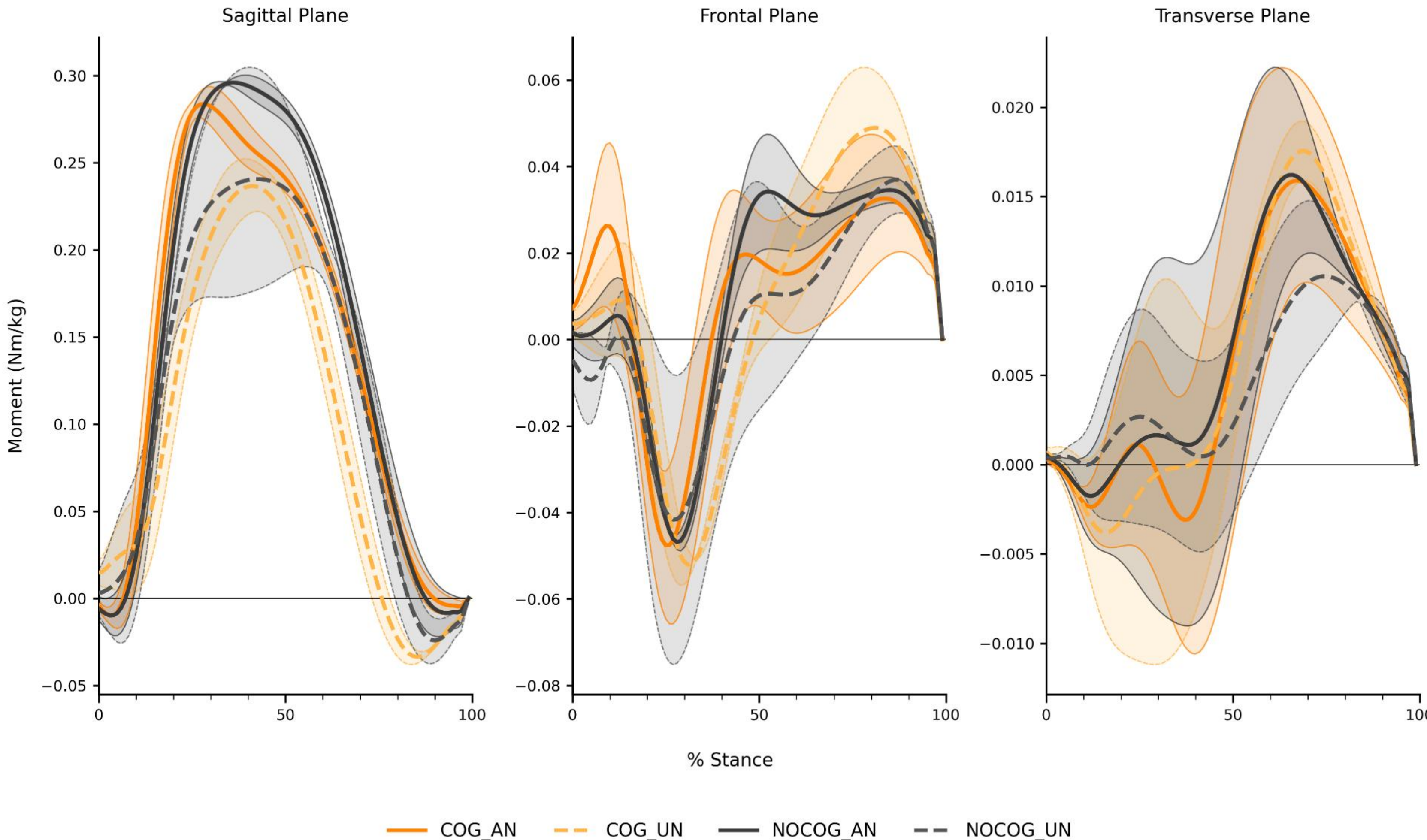

Figure 42. Subject 18 three-dimensional knee joint moments from 0 to 100% of stance phase. The shaded regions represent the standard deviation of the knee moment for each condition.

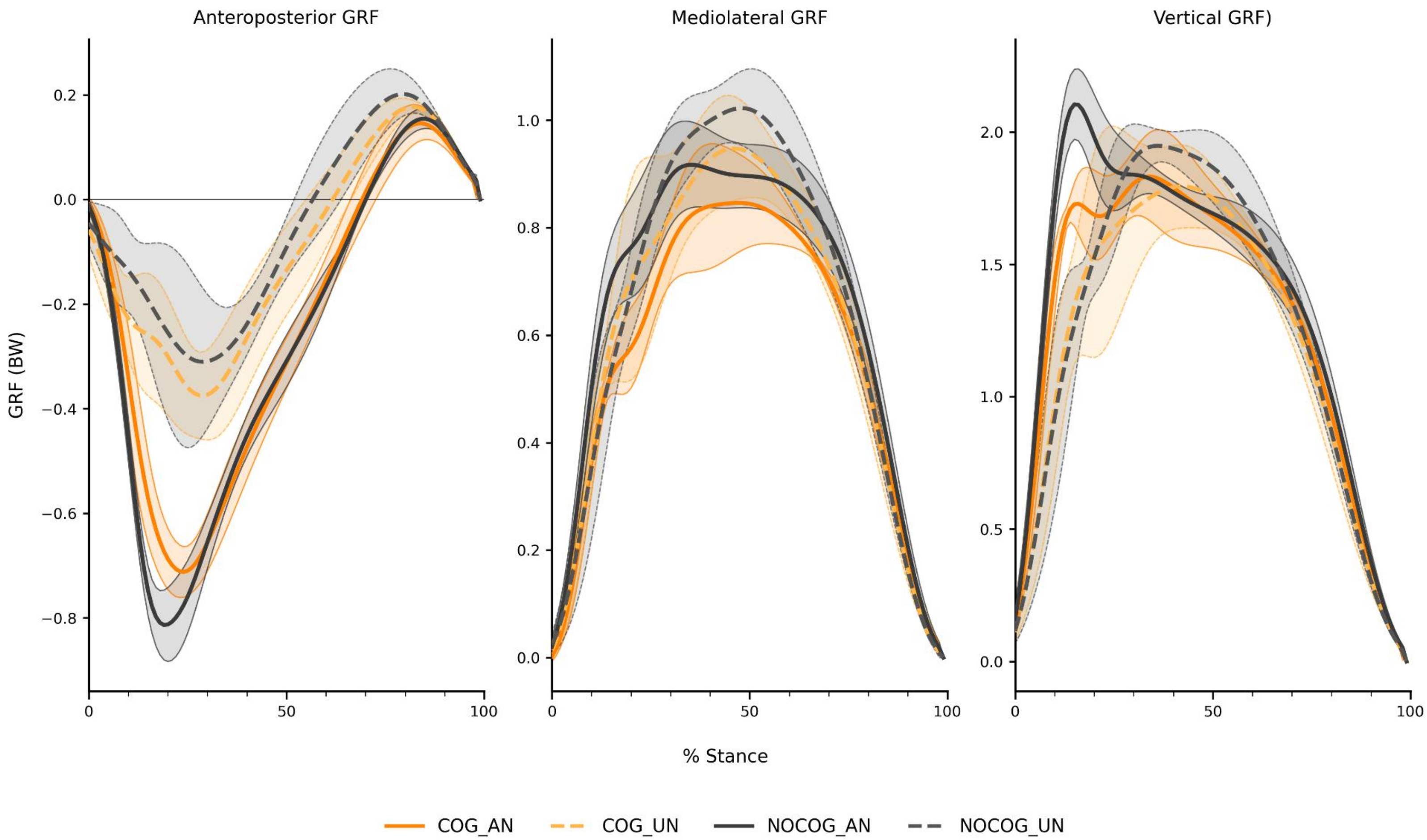

Figure 43. Subject 01 three-dimensional ground reaction forces from 0 to 100% of stance phase. The shaded regions represent the standard deviation for each condition.

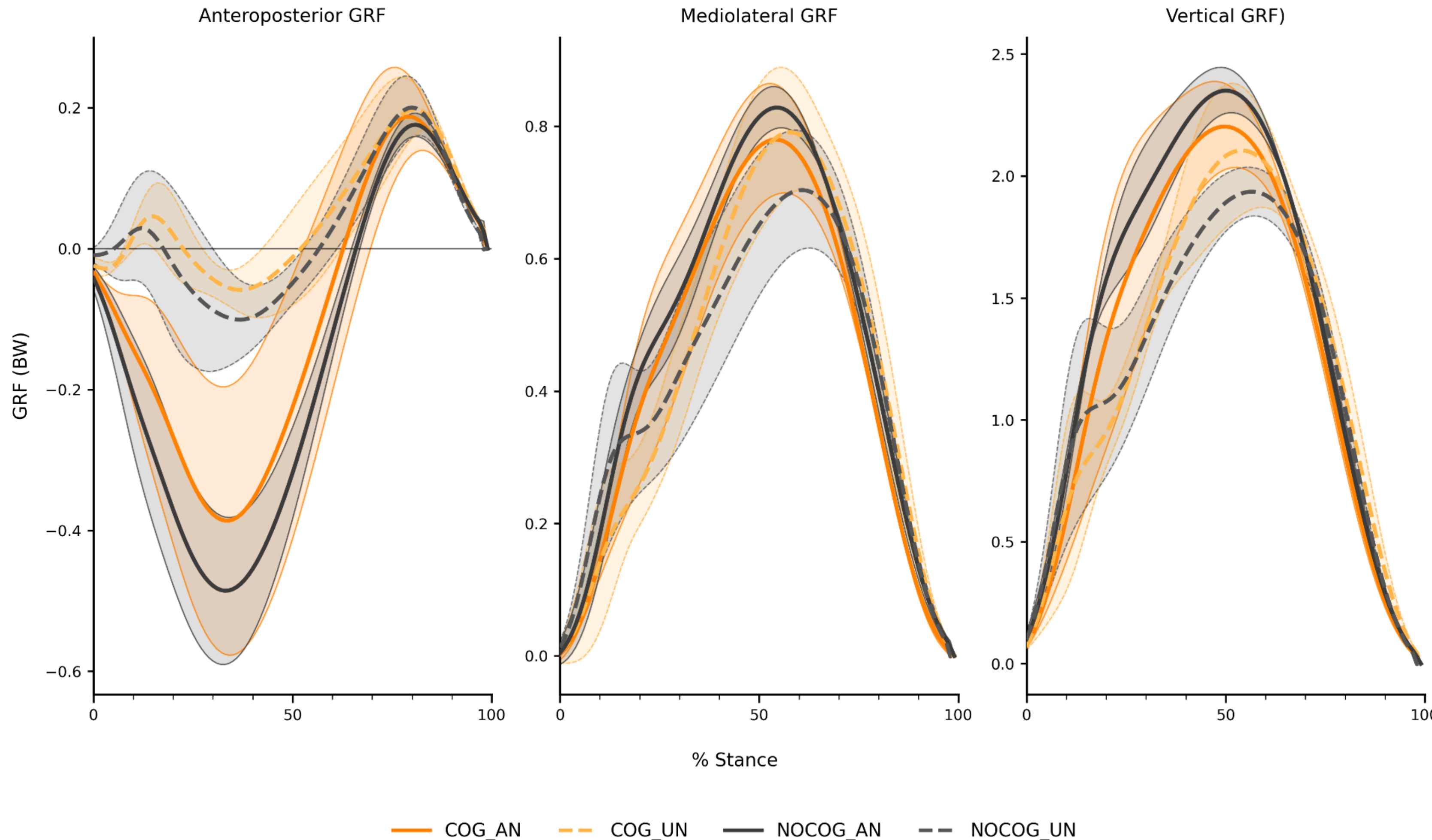

Figure 44. Subject 03 three-dimensional ground reaction forces from 0 to 100% of stance phase. The shaded regions represent the standard deviation for each condition.

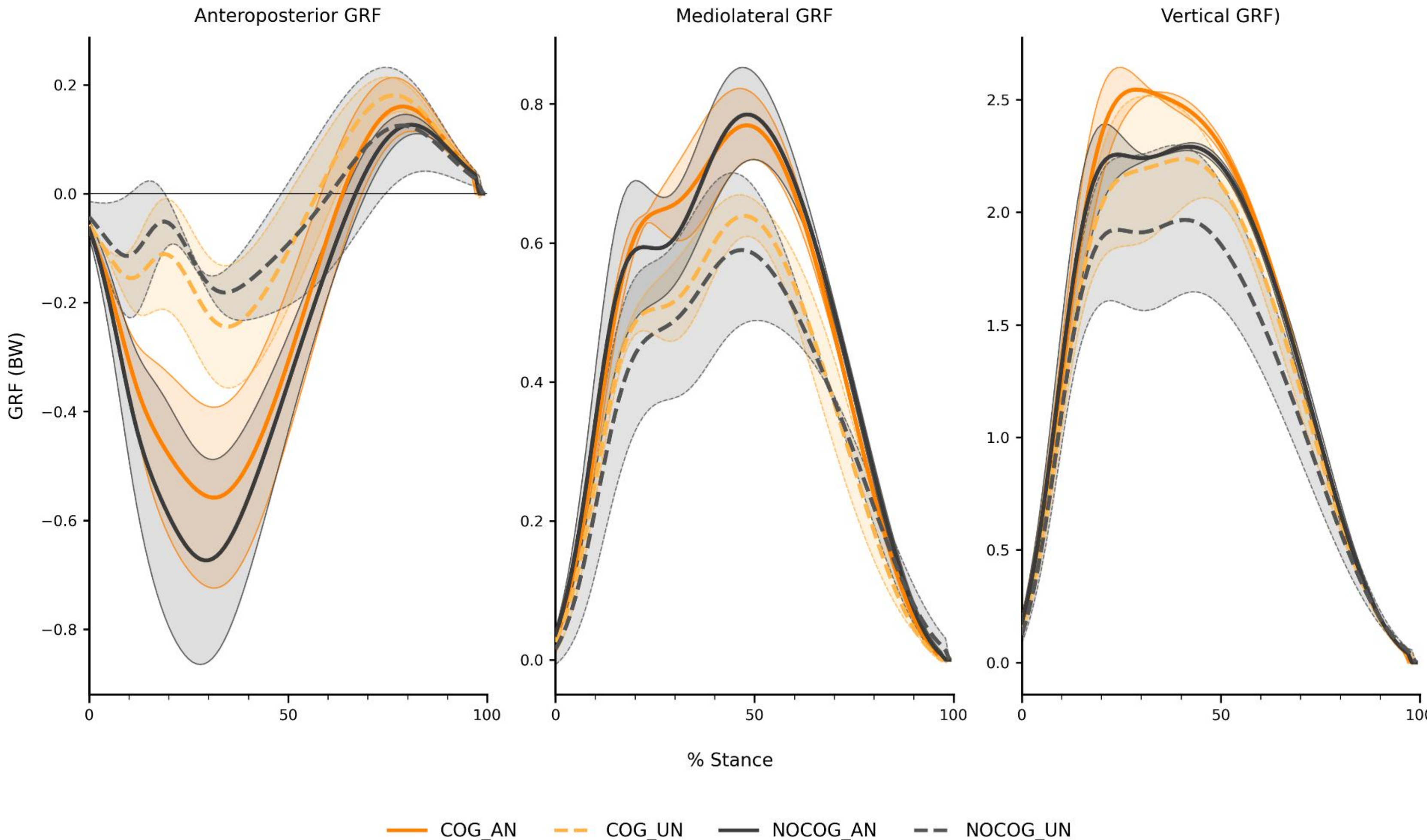

Figure 45. Subject 04 three-dimensional ground reaction forces from 0 to 100% of stance phase. The shaded regions represent the standard deviation for each condition.

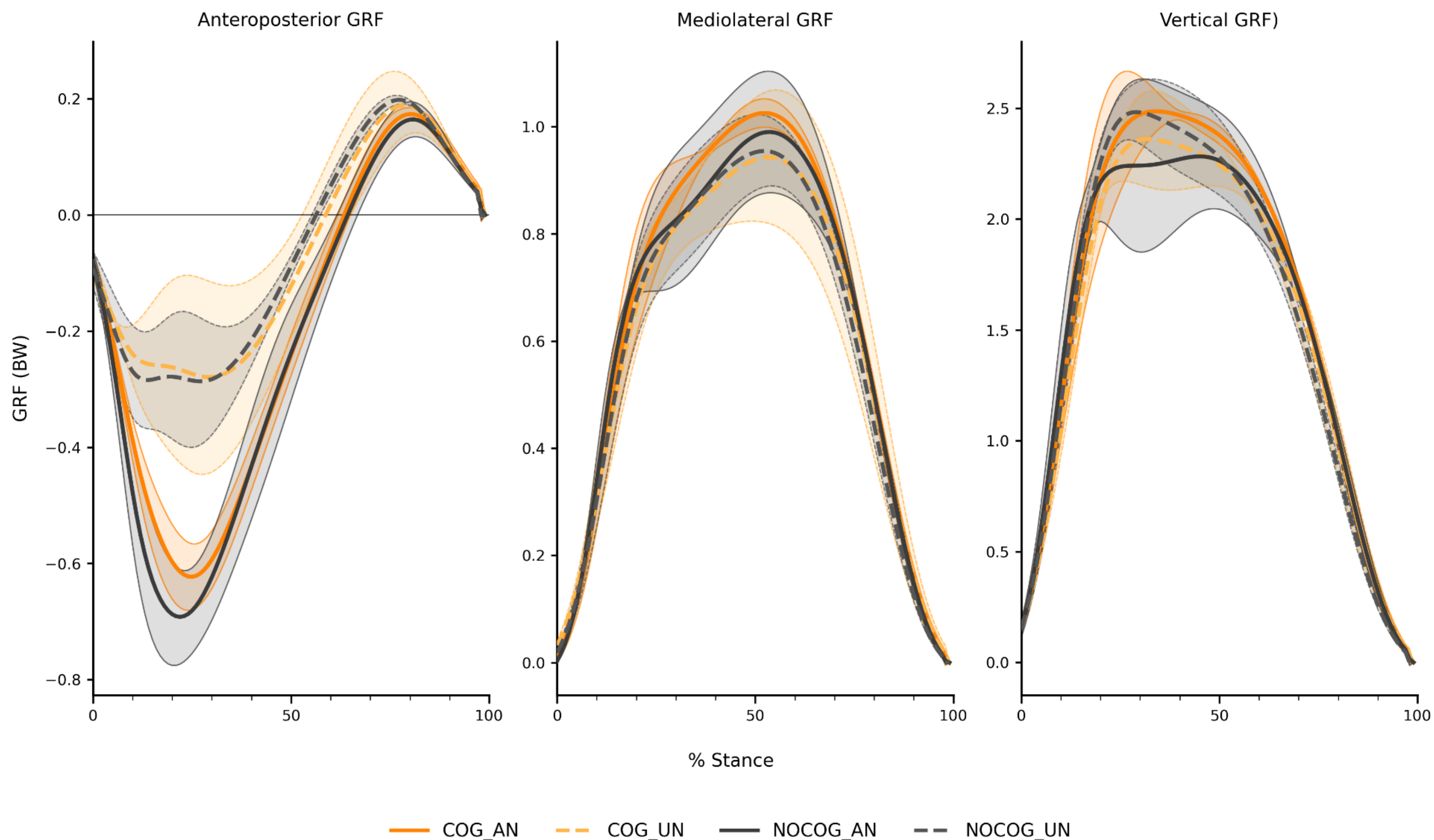

Figure 46. Subject 05 three-dimensional ground reaction forces from 0 to 100% of stance phase. The shaded regions represent the standard deviation for each condition.

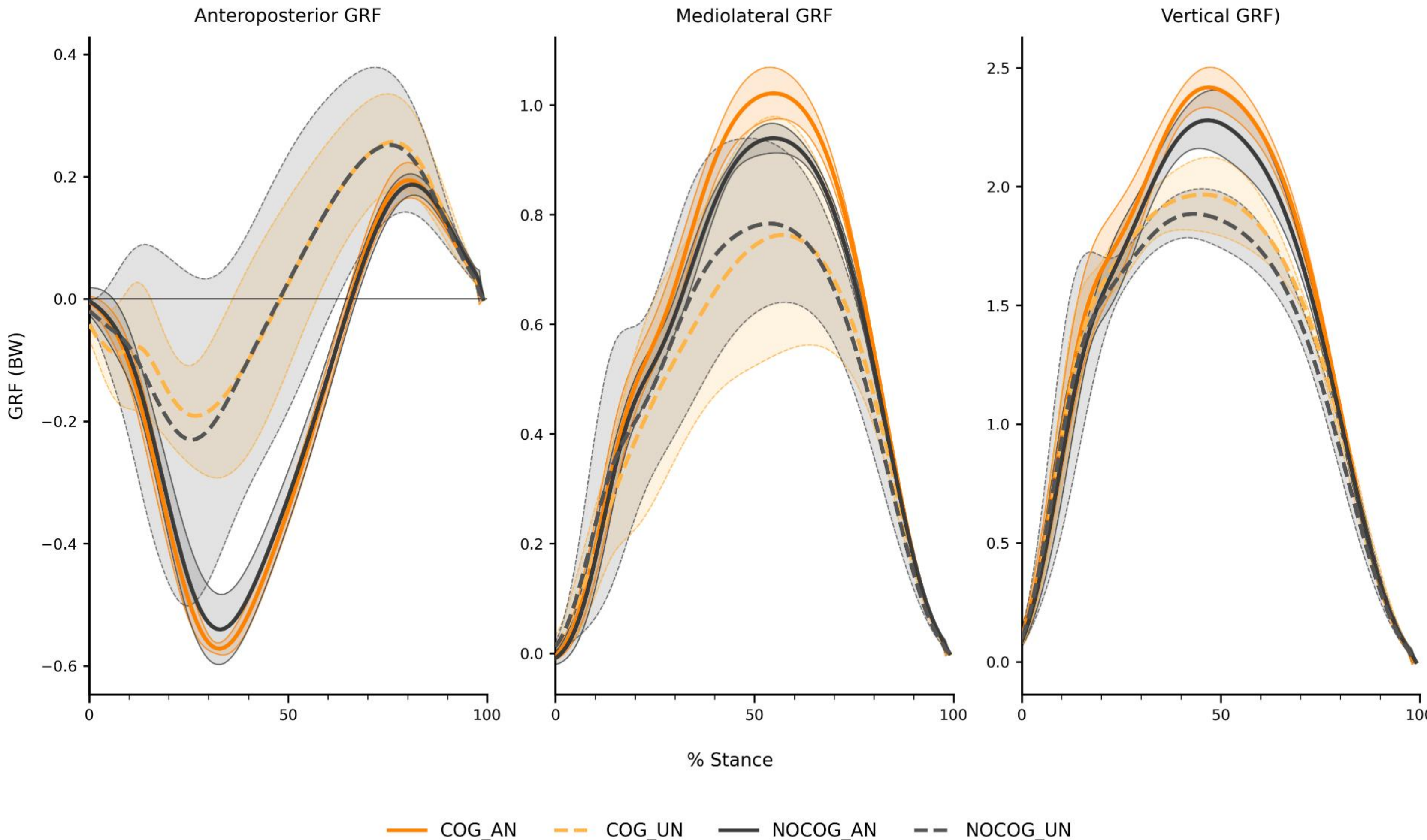

Figure 47. Subject 06 three-dimensional ground reaction forces from 0 to 100% of stance phase. The shaded regions represent the standard deviation for each condition.

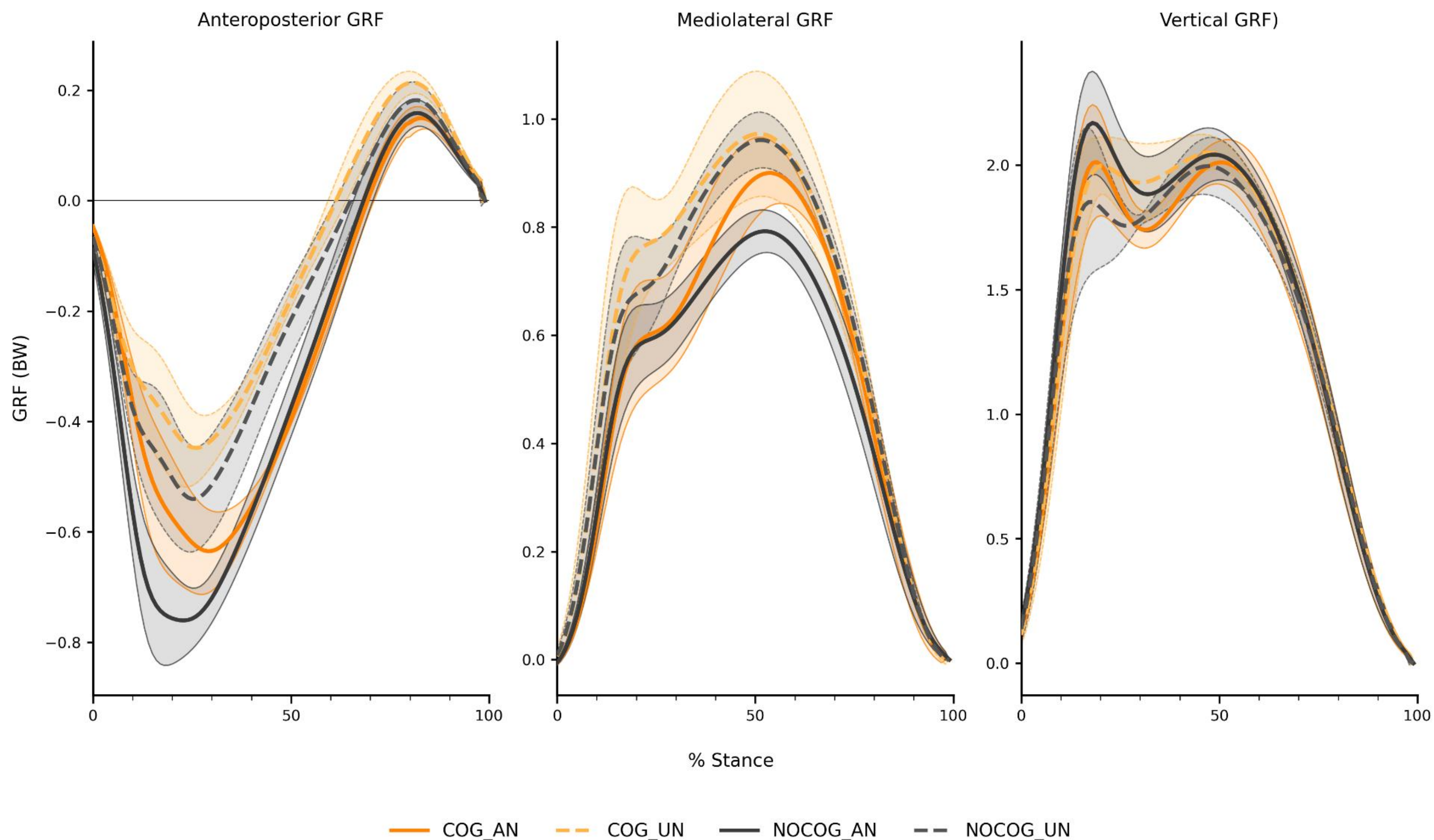

Figure 48. Subject 07 three-dimensional ground reaction forces from 0 to 100% of stance phase. The shaded regions represent the standard deviation for each condition.

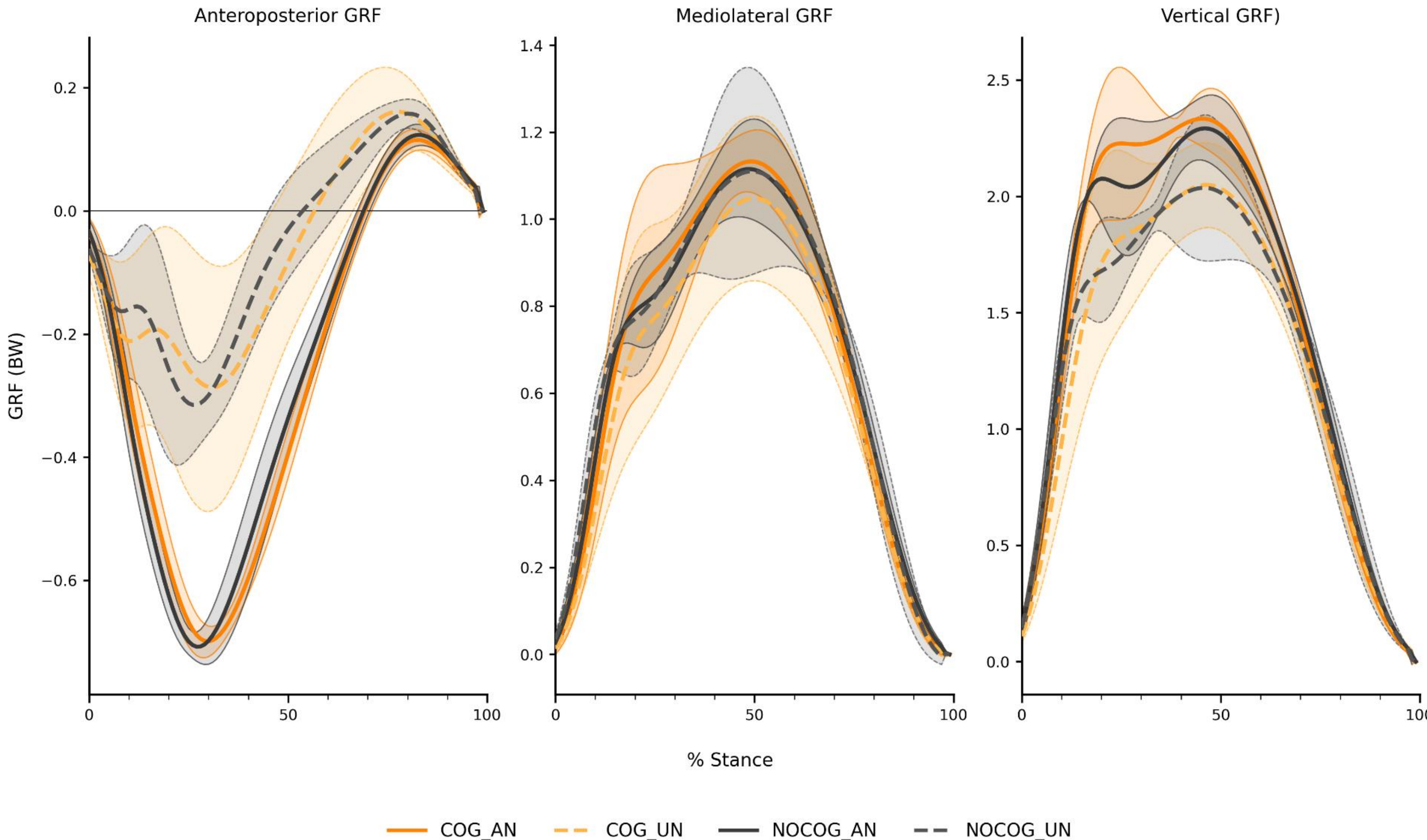

Figure 49. Subject 08 three-dimensional ground reaction forces from 0 to 100% of stance phase. The shaded regions represent the standard deviation for each condition.

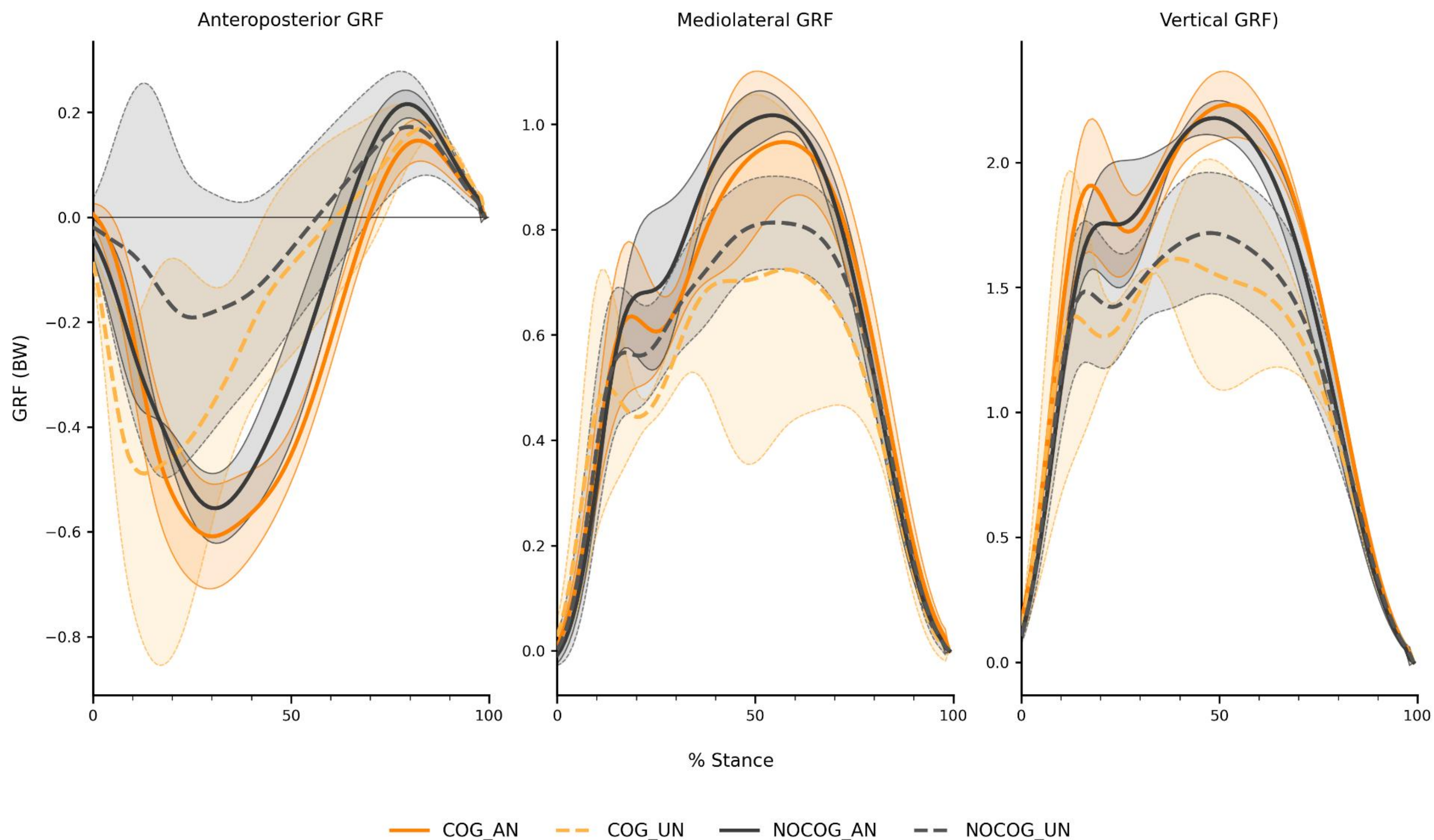

Figure 50. Subject 09 three-dimensional ground reaction forces from 0 to 100% of stance phase. The shaded regions represent the standard deviation for each condition.

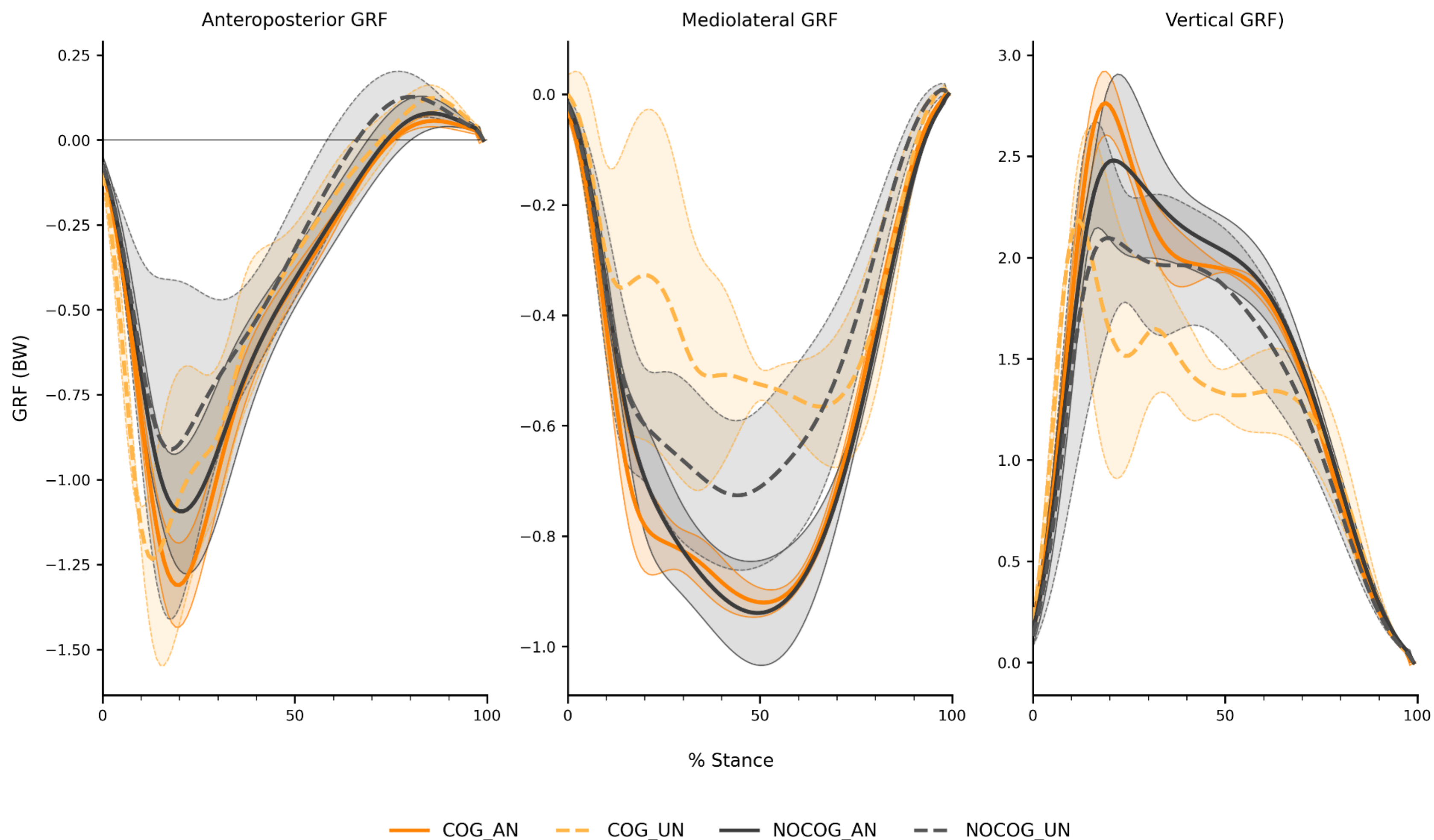

Figure 51. Subject 10 three-dimensional ground reaction forces from 0 to 100% of stance phase. The shaded regions represent the standard deviation for each condition.

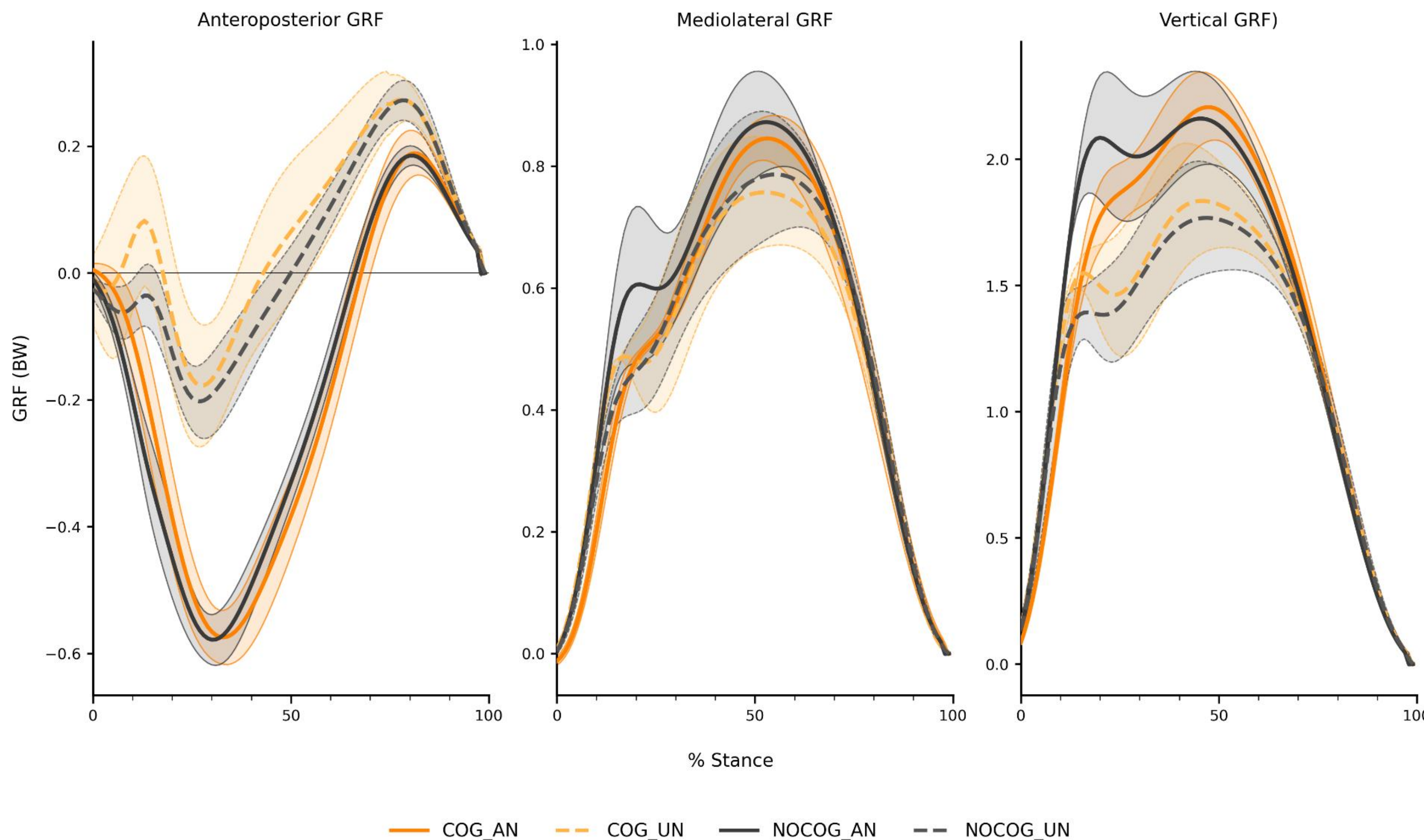

Figure 52. Subject 11 three-dimensional ground reaction forces from 0 to 100% of stance phase. The shaded regions represent the standard deviation for each condition.

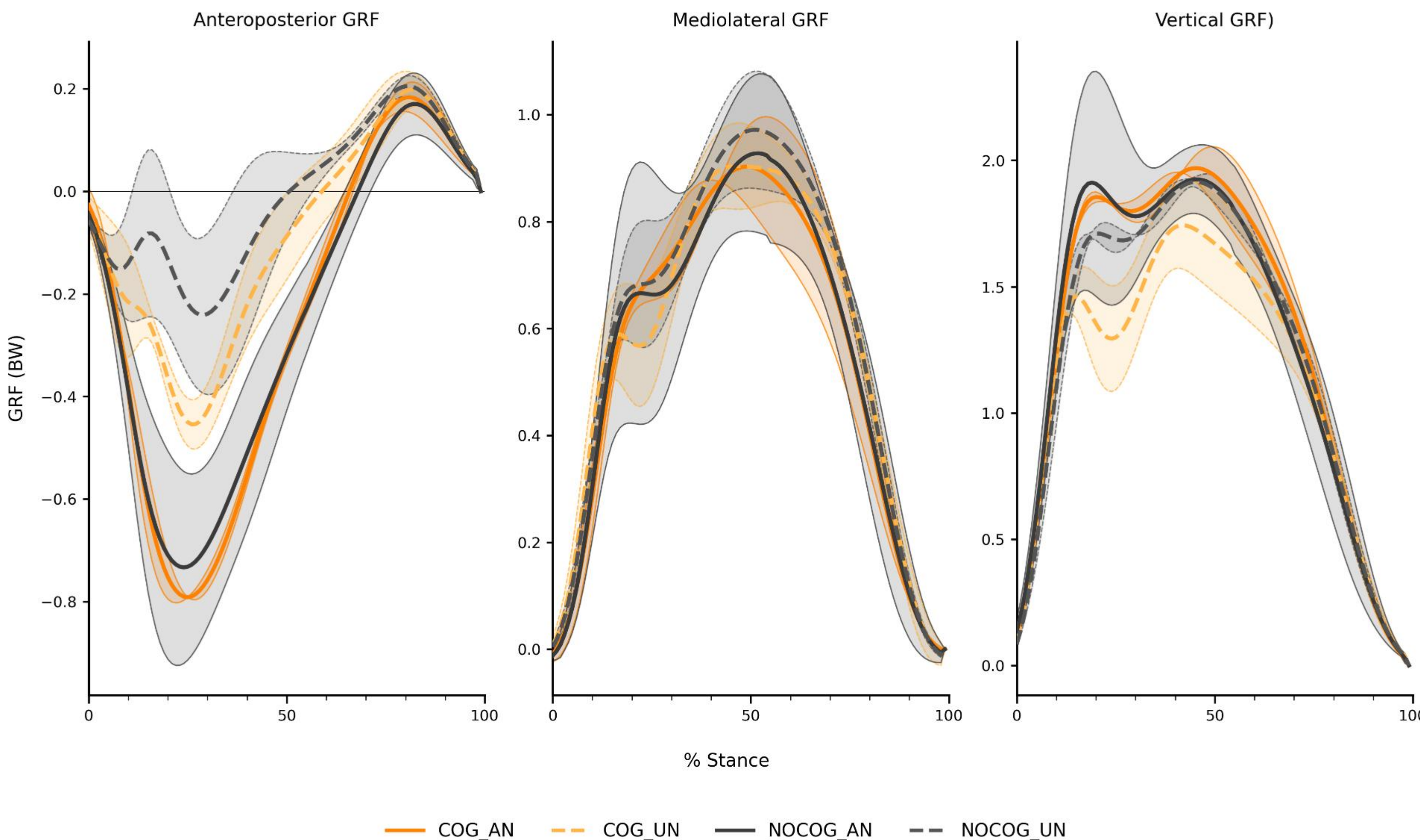

Figure 53. Subject 12 three-dimensional ground reaction forces from 0 to 100% of stance phase. The shaded regions represent the standard deviation for each condition.

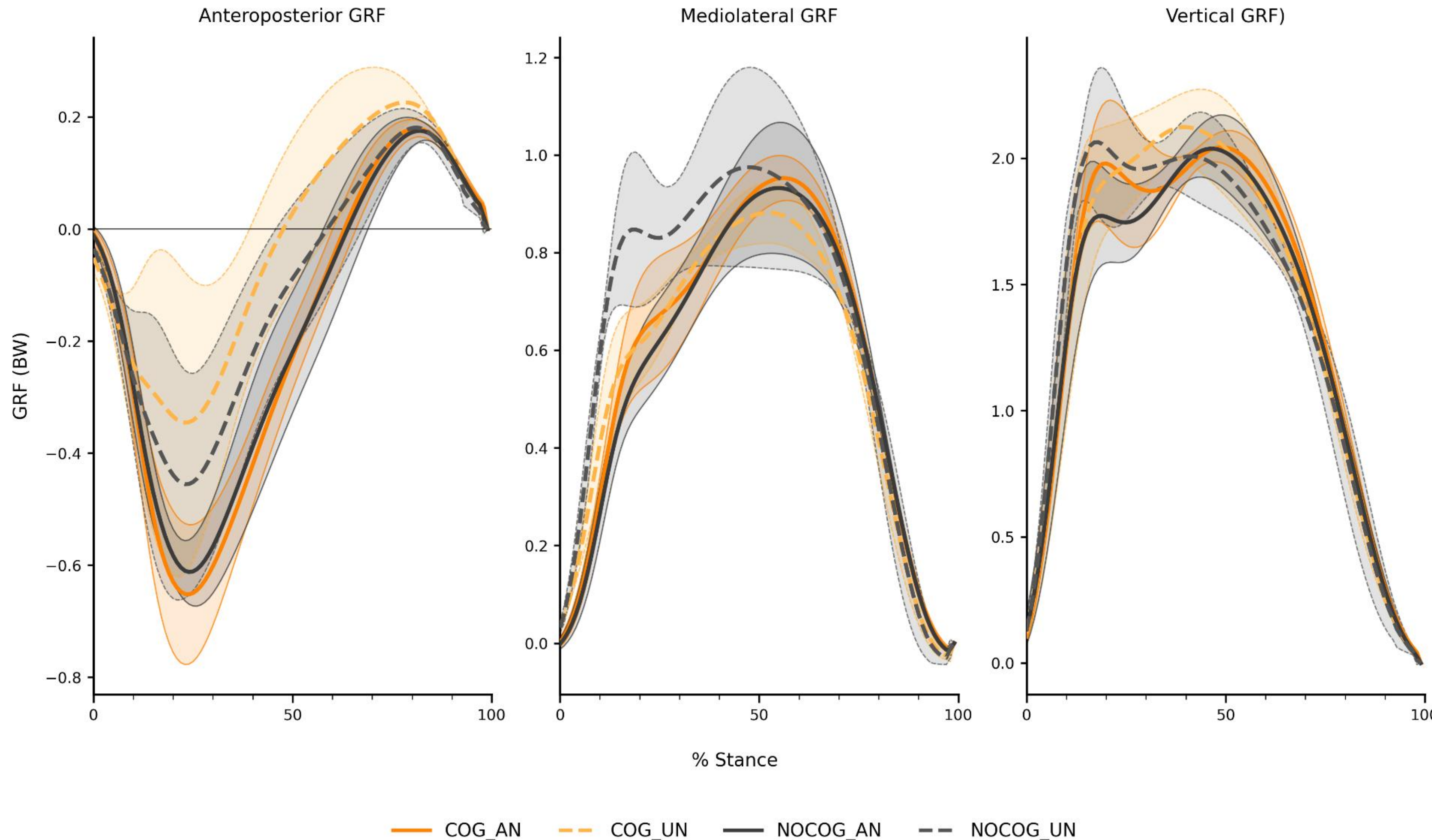

Figure 54. Subject 13 three-dimensional ground reaction forces from 0 to 100% of stance phase. The shaded regions represent the standard deviation for each condition.

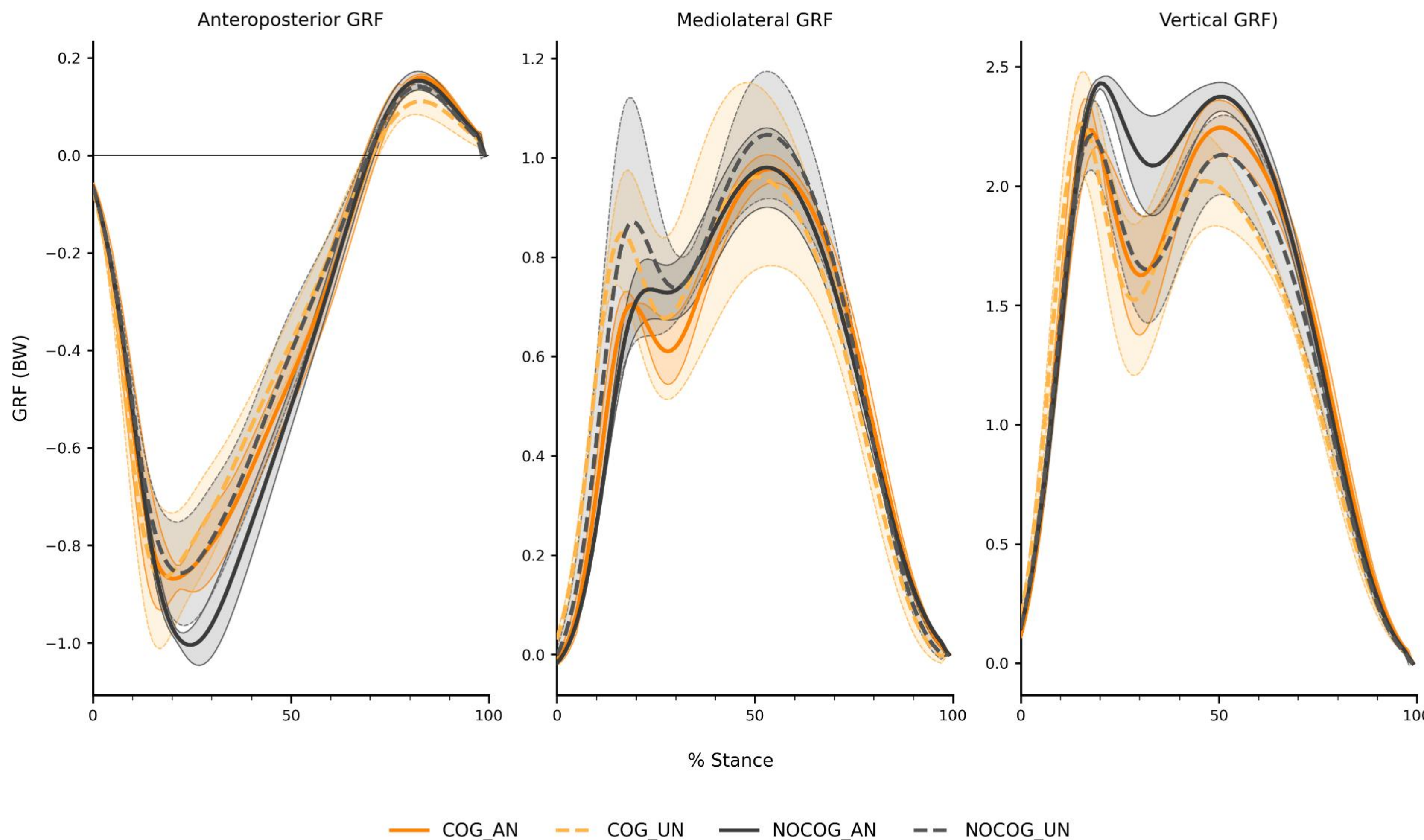

Figure 55. Subject 14 three-dimensional ground reaction forces from 0 to 100% of stance phase. The shaded regions represent the standard deviation for each condition.

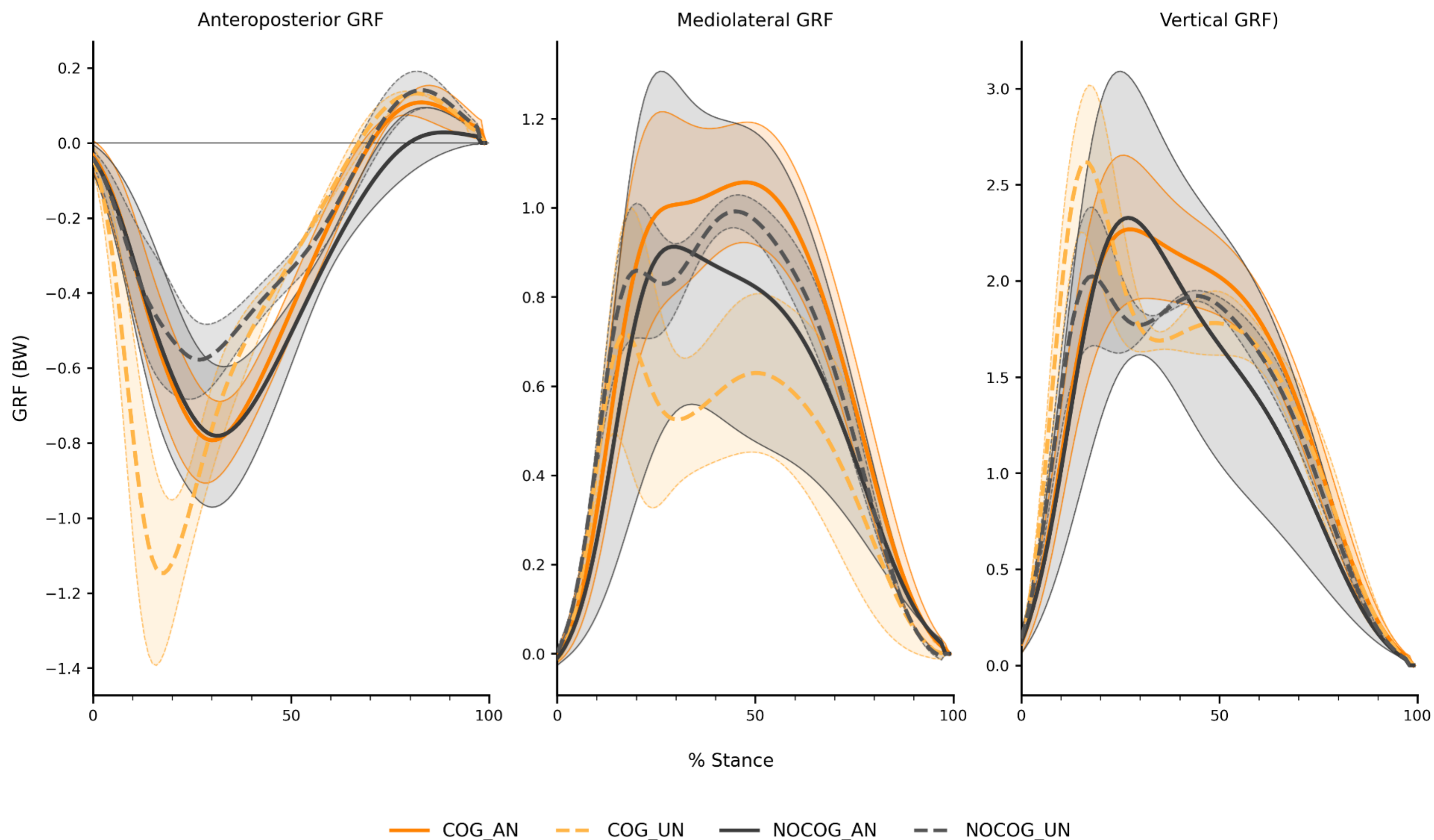

Figure 56. Subject 15 three-dimensional ground reaction forces from 0 to 100% of stance phase. The shaded regions represent the standard deviation for each condition.

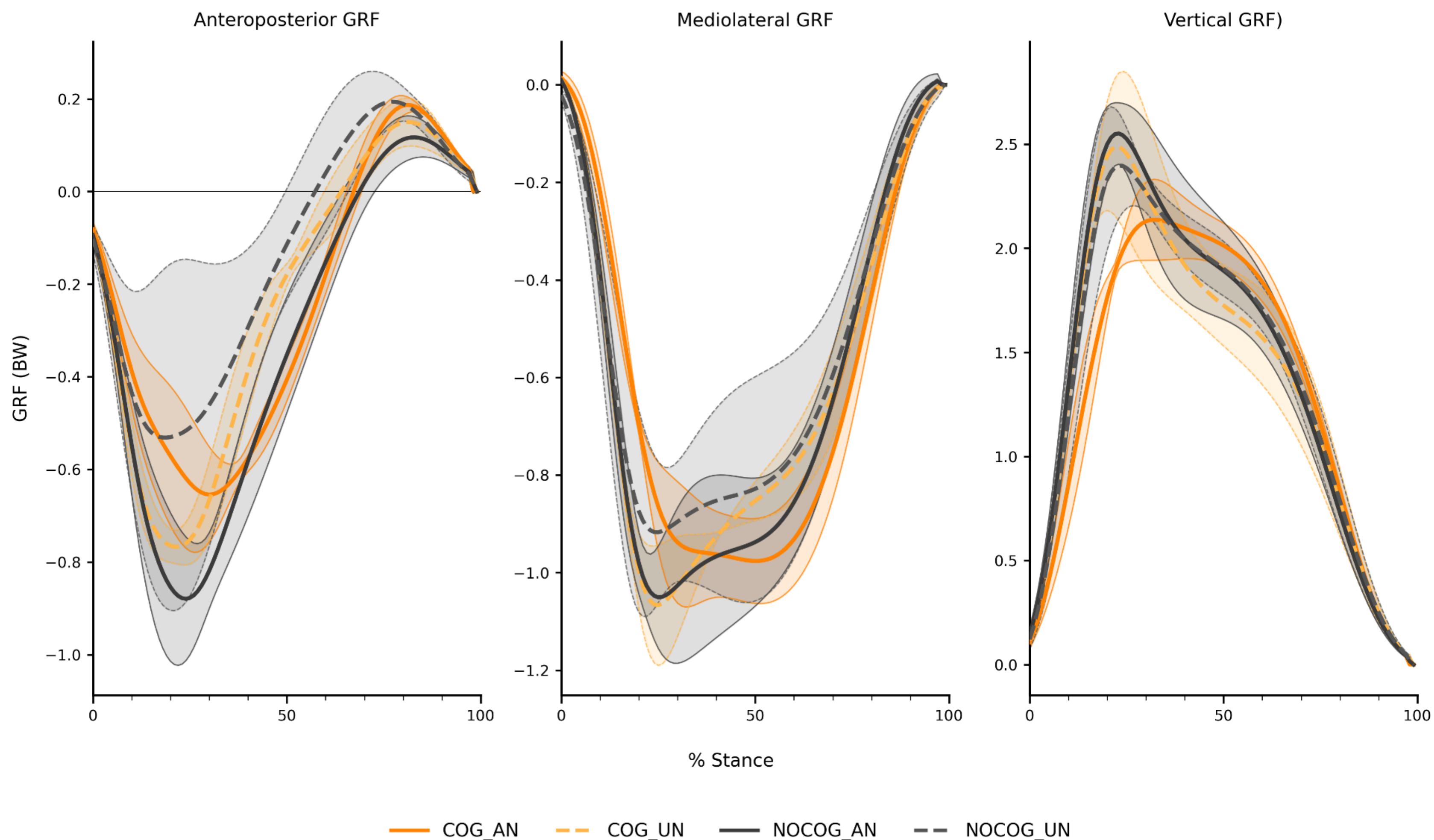

Figure 57. Subject 16 three-dimensional ground reaction forces from 0 to 100% of stance phase. The shaded regions represent the standard deviation for each condition.

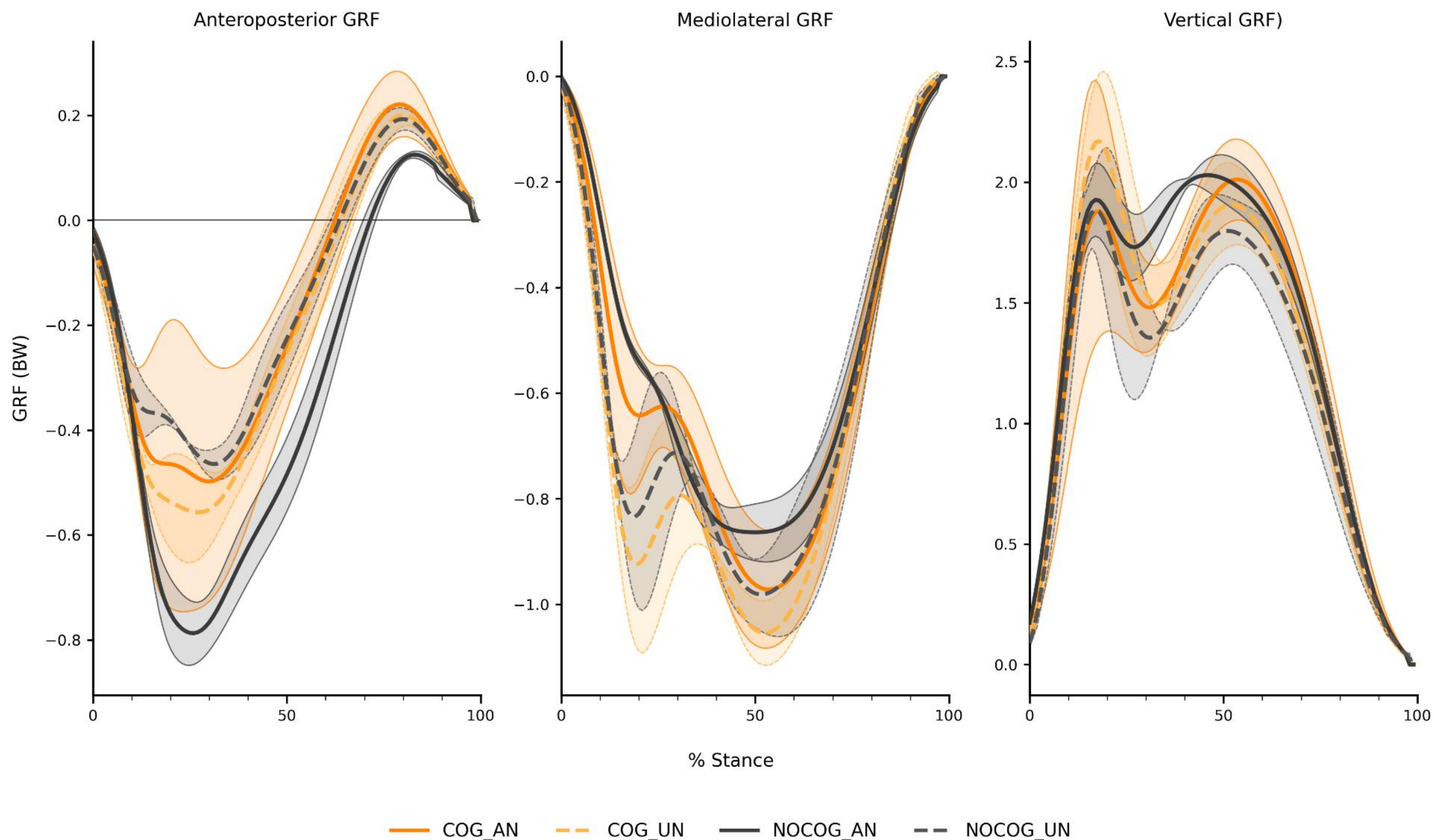

Figure 58. Subject 17 three-dimensional ground reaction forces from 0 to 100% of stance phase. The shaded regions represent the standard deviation for each condition.

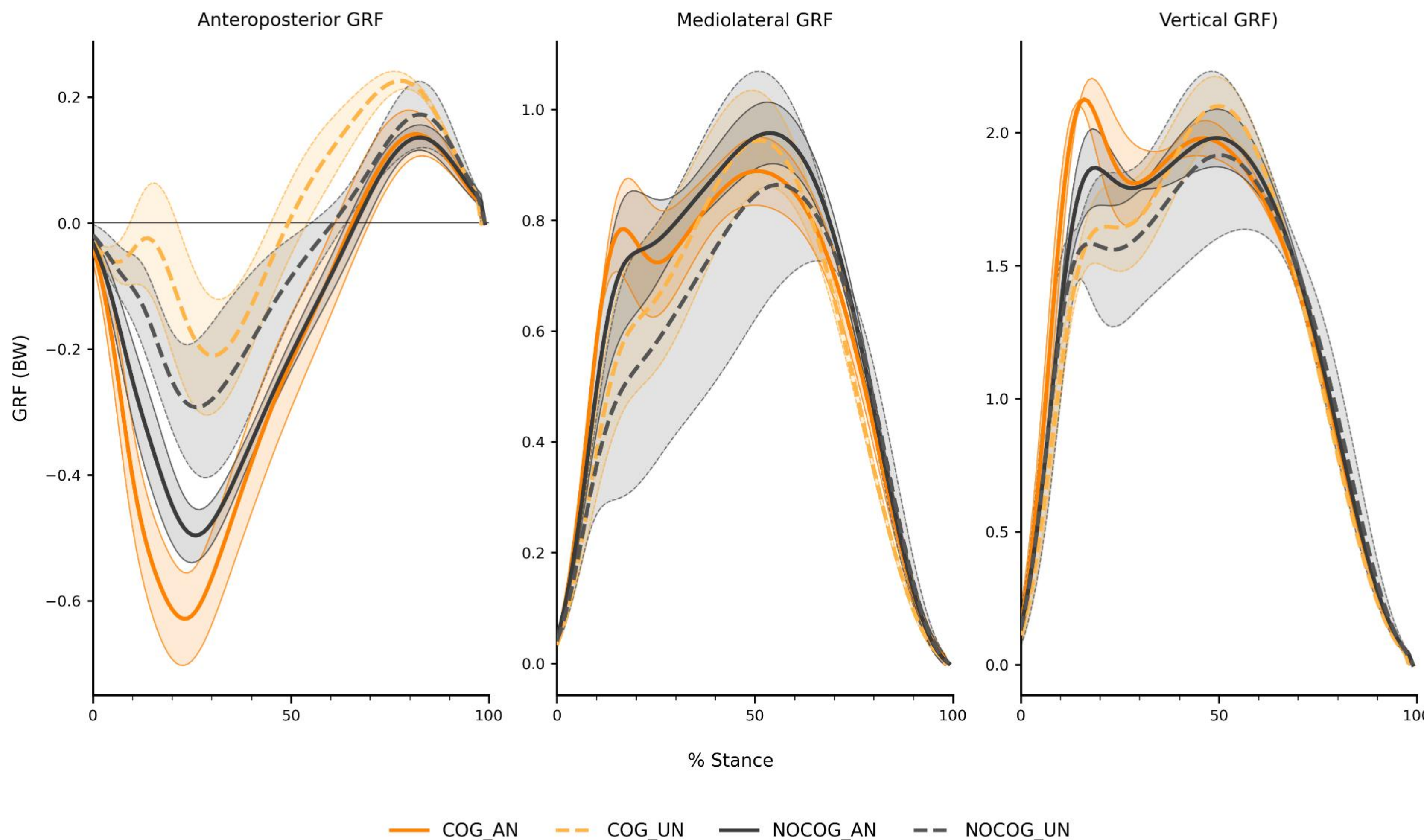

Figure 59. Subject 18 three-dimensional ground reaction forces from 0 to 100% of stance phase. The shaded regions represent the standard deviation for each condition.

Table 22. Fixed-effect results from the linear mixed-effects model on initial contact knee flexion angle.

Variable	**Term**	**β**	**SE**	***z***	***p***	**95% CI [Lower, Upper]**
Knee Flexion (deg) Initial Contact	Intercept: NOCOG_AN	-28.8	2.4	-12.1	< 0.001	[-33.5, -24.1]
	Cognitive Load: COG vs NOCOG	1.1	1.9	0.6	0.570	[-2.6, 4.7]
	Anticipation: UN vs AN	-13.8	1.9	-7.2	**< 0.001**	[-17.5, -10.1]
	Interaction: COG x AN	0.8	2.6	0.3	0.764	[-4.3, 5.9]
	Entrance Time	1.3	36.8	0.0	0.972	[-70.8, 73.4]

Note. β = unstandardized coefficient; SE = standard error; CI = 95% confidence interval. The intercept represents the predicted mean for the reference condition (NOCOG_AN) and is not interpreted for significance. Bold values indicate $p < .05$.

Table 23. Fixed-effect results from the linear mixed-effects model on initial contact knee abduction angle.

Variable	Term	β	SE	z	p	95% CI [Lower, Upper]
Knee Abduction (deg) Initial Contact	Intercept: NOCOG_AN	-1.1	0.8	-1.4	0.15	[-2.57, 0.41]
	Cognitive Load: COG vs NOCOG	0.0	0.5	-0.1	0.95	[-0.92, 0.86]
	Anticipation: UN vs AN	-0.7	0.5	-1.6	0.12	[-1.64, 0.18]
	Interaction: (COG vs NOCOG) × (UN vs AN)	-0.3	0.6	-0.4	0.66	[-1.53, 0.97]
	Entrance Time	-4.3	9.0	-0.5	0.63	[-21.84, 13.29]

Note. β = unstandardized coefficient; SE = standard error; CI = 95% confidence interval. The intercept represents the predicted mean for the reference condition (NOCOG_AN) and is not interpreted for significance. Bold values indicate $p < .05$.

Table 24. Fixed-effect results from the linear mixed-effects model on initial contact knee external/internal rotation angle.

Variable	Term	β	SE	*z*	*p*	95% CI [Lower, Upper]
External/Internal Rotation Initial Contact	Intercept: NOCOG_AN	-28.8	2.4	-12.1	< 0.001	[33.50, -24.14]
	Cognitive Load: COG vs NOCOG	1.1	1.9	0.6	0.571	[-2.60, 4.71]
	Anticipation: UN vs AN	-13.8	1.9	-7.2	**< 0.001**	[-17.52, -10.06]
	Interaction: (COG vs NOCOG) × (UN vs AN)	0.8	2.6	0.3	0.764	[-4.34, 5.91]
	Entrance Time	1.3	36.8	0.0	0.972	[-70.78, 73.39]

Note. β = unstandardized coefficient; SE = standard error; CI = 95% confidence interval. The intercept represents the predicted mean for the reference condition (NOCOG_AN) and is not interpreted for significance. Bold values indicate $p < .05$.

Table 25. Fixed-effect results from the linear mixed-effects model on peak knee flexion angle.

Variable	Term	β	SE	*z*	*p*	95% CI [Lower, Upper]
Peak Flexion Angle (deg)	Intercept: NOCOG_AN	-58.2	1.7	-35.2	< 0.001	[-61.47, -54.98]
	Cognitive Load: COG vs NOCOG	0.8	0.9	0.9	0.379	[-1.01, 2.66]
	Anticipation: UN vs AN	-1.9	1.0	-2.0	**0.042**	[-3.82, -0.07]
	Interaction: (COG vs NOCOG) × (UN vs AN)	-1.1	1.3	-0.8	0.414	[-3.64, 1.50]
	Entrance Time	14.3	18.4	0.8	0.439	[-21.88, 50.44]

Note. β = unstandardized coefficient; SE = standard error; CI = 95% confidence interval. The intercept represents the predicted mean for the reference condition (NOCOG_AN) and is not interpreted for significance. Bold values indicate $p < .05$.

Table 26. Fixed-effect results from the linear mixed-effects model on peak knee adduction/abduction angle.

Variable	Term	β	SE	*z*	*p*	95% CI [Lower, Upper]
Peak Adduction Angle (deg)	Intercept: NOCOG_AN	-7.9	0.9	-8.4	< 0.001	[-9.78, -6.08]
	Cognitive Load: COG vs NOCOG	0.0	0.5	-0.1	0.924	[-0.99, 0.90]
	Anticipation: UN vs AN	-0.4	0.5	-0.8	0.435	[-1.34, 0.58]
	Interaction: (COG vs NOCOG) × (UN vs AN)	0.2	0.7	0.3	0.760	[-1.11, 1.53]
	Entrance Time	4.1	9.5	0.4	0.669	[-14.51, 22.62]

Note. β = unstandardized coefficient; SE = standard error; CI = 95% confidence interval. The intercept represents the predicted mean for the reference condition (NOCOG_AN) and is not interpreted for significance. Bold values indicate $p < .05$.

Table 27. Fixed-effect results from the linear mixed-effects model on peak knee external rotation angle.

Variable	Term	β	SE	*z*	*p*	95% CI [Lower, Upper]
Peak External Rotation (deg)	Intercept: NOCOG_AN	13.3	1.0	13.4	< 0.001	[11.53, 15.22]
	Cognitive Load: COG vs NOCOG	0.07	0.38	0.17	0.862	[-0.68, 0.81]
	Anticipation: UN vs AN	-0.06	0.39	-0.17	0.868	[-0.82, 0.70]
	Interaction: (COG vs NOCOG) × (UN vs AN)	-0.12	0.53	-0.23	0.816	[-1.17, 0.92]
	Entrance Time	-0.68	7.49	-0.09	0.928	[-15.35, 14.00]

Note. β = unstandardized coefficient; SE = standard error; CI = 95% confidence interval. The intercept represents the predicted mean for the reference condition (NOCOG_AN) and is not interpreted for significance. Bold values indicate $p < .05$.

Table 28. Fixed-effect results from the linear mixed-effects model on peak knee extension moment.

Variable	Term	β	SE	*z*	*p*	95% CI [Lower, Upper]
Peak Extension (Nm/kg)	Intercept: NOCOG_AN	3.39	0.12	29.16	< 0.001	[3.16, 3.62]
	Cognitive Load: COG vs NOCOG	-0.06	0.08	-0.75	0.455	[-0.22, 0.10]
	Anticipation: UN vs AN	-0.67	0.08	-8.08	**< 0.001**	[-0.84, -0.51]
	Interaction: (COG vs NOCOG) × (UN vs AN)	0.22	0.11	1.90	0.057	[-0.01, 0.44]
	Entrance Time	0.04	1.62	0.03	0.978	[-3.13, 3.22]

Note. β = unstandardized coefficient; SE = standard error; CI = 95% confidence interval. The intercept represents the predicted mean for the reference condition (NOCOG_AN) and is not interpreted for significance. Bold values indicate $p < .05$.

Table 29. Fixed-effect results from the linear mixed-effects model on peak knee adduction moment.

Variable	Term	β	SE	*z*	*p*	95% CI [Lower, Upper]
Peak Knee Adduction Moment (Nm/kg)	Intercept: NOCOG_AN	0.48	0.07	7.26	< 0.001	[0.35, 0.61]
	Cognitive Load: COG vs NOCOG	0.01	0.04	0.14	0.892	[-0.07, 0.08]
	Anticipation: UN vs AN	-0.10	0.04	-2.59	0.009	[-0.18, -0.03]
	Interaction: (COG vs NOCOG) × (UN vs AN)	0.03	0.05	0.57	0.567	[-0.08, 0.14]
	Entrance Time	-0.85	0.77	-1.10	0.273	[-2.36, 0.67]

Note. β = unstandardized coefficient; SE = standard error; CI = 95% confidence interval. The intercept represents the predicted mean for the reference condition (NOCOG_AN) and is not interpreted for significance. Bold values indicate $p < .05$.

Table 30. Fixed-effect results from the linear mixed-effects model on peak knee external rotation moment.

Variable	Term	β	SE	*z*	*p*	95% CI [Lower, Upper]
Peak External Rotation (Nm/kg)	Intercept: NOCOG_AN	-0.18	0.06	-3.22	0.001	[-0.29, -0.07]
	Cognitive Load: COG vs NOCOG	0.01	0.06	0.19	0.847	[-0.11, 0.13]
	Anticipation: UN vs AN	0.01	0.06	0.23	0.818	[0.11, 0.14]
	Interaction: (COG vs NOCOG) × (UN vs AN)	-0.09	0.09	-1.03	0.302	[-0.26, 0.08]
	Entrance Time	-0.05	1.21	-0.04	0.967	[-2.42, 2.31]

Note. β = unstandardized coefficient; SE = standard error; CI = 95% confidence interval. The intercept represents the predicted mean for the reference condition (NOCOG_AN) and is not interpreted for significance. Bold values indicate $p < .05$.

Table 31. Fixed-effect results from the linear mixed-effects model on peak vertical ground reaction force.

Variable	Term	β	SE	z	p	95% CI [Lower, Upper]
	Intercept: NOCOG_AN	2.28	0.05	46.21	0.000	[2.18, 2.37]
	Cognitive Load: COG vs NOCOG	-0.01	0.04	-0.17	0.864	[-0.09, 0.07]
Vertical GRF (BW)	Anticipation: UN vs AN	-0.22	0.04	-5.08	**< 0.001**	[-0.30, -0.13]
	Interaction: (COG vs NOCOG) × (UN vs AN)	0.09	0.06	1.47	0.141	[0.03, 0.20]
	Entrance Time	0.86	0.82	1.05	0.294	[-0.75, 2.48]

Note. β = unstandardized coefficient; SE = standard error; CI = 95% confidence interval. The intercept represents the predicted mean for the reference condition (NOCOG_AN) and is not interpreted for significance. Bold values indicate $p < .05$.

Table 32. Fixed-effect results from the linear mixed-effects model on peak anteroposterior ground reaction force.

Variable	Term	β	SE	*z*	*p*	95% CI [Lower, Upper]
Anteroposterior GRF (BW)	Intercept: NOCOG_AN	-0.72	0.06	-12.95	< 0.001	[-0.83, -0.61]
	Cognitive Load: COG vs NOCOG	0.04	0.03	1.14	0.253	[-0.03, 0.10]
	Anticipation: UN vs AN	0.31	0.03	9.63	**< 0.000**	[0.25, 0.37]
	Interaction: (COG vs NOCOG) × (UN vs AN)	-0.12	0.04	-2.73	**0.006**	[-0.21, -0.03]
	Entrance Time	0.34	0.62	0.55	0.583	[-0.87, 1.55]

Note. β = unstandardized coefficient; SE = standard error; CI = 95% confidence interval. The intercept represents the predicted mean for the reference condition (NOCOG_AN) and is not interpreted for significance. Bold values indicate *p* < .05.

Table 31.B. Post-hoc pairwise comparisons from the linear mixed-effects model on peak anteroposterior ground reaction force.

Contrast	Δβ	SE	*t*	*p adjusted*	95% CI [Lower, Upper]	Correction
NOCOG_AN vs NOCOG_UN	-0.308	0.032	-9.630	0.000	[-0.37, -0.27]	Bonferroni
NOCOG_AN vs COG_AN	-0.036	0.031	-1.143	1.000	[-0.09, 0.03]	
NOCOG_AN vs COG_UN	-0.224	0.032	-7.043	0.000	[-0.29, -0.16]	
NOCOG_UN vs COG_AN	0.272	0.032	8.597	0.000	[0.21, 0.34]	
NOCOG_UN vs COG_UN	0.085	0.031	2.754	0.038	[0.02, 0.15]	
COG_AN vs COG_UN	-0.188	0.032	-5.936	0.000	[-0.25, -0.13]	

Note. Δβ = estimated mean difference between conditions; SE = standard error; CI = 95% confidence interval; *p-adjusted* = Bonferroni-corrected *p* value. Comparisons are pairwise contrasts of the estimated marginal means from the mixed-effects model. Bold values indicate *p* < .05.

Table 33. Fixed-effect results from the linear mixed-effects model on peak mediolateral ground reaction force.

Variable	Term	β	SE	*z*	*p*	95% CI [Lower, Upper]
Mediolateral GRF (BW)	Intercept: NOCOG_AN	0.58	0.17	3.36	< 0.001	[0.24, 0.93]
	Cognitive Load: COG vs NOCOG	0.01	0.02	0.58	0.563	[-0.03, 0.06]
	Anticipation: UN vs AN	0.00	0.02	0.12	0.905	[-0.04, 0.05]
	Interaction: (COG vs NOCOG) × (UN vs AN)	-0.05	0.03	-1.56	0.118	[-0.12, 0.01]
	Entrance Time	-0.63	0.46	-1.36	0.173	[-1.54, 0.28]

Note. β = unstandardized coefficient; SE = standard error; CI = 95% confidence interval. The intercept represents the predicted mean for the reference condition (NOCOG_AN) and is not interpreted for significance. Bold values indicate $p < .05$.

Table 34. Fixed-effect results from the linear mixed-effects model on medial hamstring activation amplitude prior to initial contact.

Variable	Term	β	SE	*z*	*p*	95% CI [Lower, Upper]
Medial Hamstring	Intercept: NOCOG_AN	0.79	0.09	8.94	0.000	[0.62, 0.97]
	Cognitive Load: COG vs NOCOG	-0.03	0.04	-0.71	0.477	[-0.12, 0.05]
	Anticipation: UN vs AN	-0.10	0.04	-2.14	**0.032**	[-0.18, -0.01]
	Interaction: (COG vs NOCOG) × (UN vs AN)	0.10	0.06	1.59	0.111	[-0.02, 0.22]
	Entrance Time	-0.68	0.86	-0.79	0.432	[-2.37, 1.02]

Note. β = unstandardized coefficient; SE = standard error; CI = 95% confidence interval. The intercept represents the predicted mean for the reference condition (NOCOG_AN) and is not interpreted for significance. Pre-initial contact EMG amplitudes were calculated as the mean activation within the 50 ms window preceding force plate contact for the dominant limb. Bold values indicate $p < .05$.

Table 35. Fixed-effect results from the linear mixed-effects model on medial hamstring activation amplitude 100 ms post initial contact.

Variable	Term	β	SE	*z*	*p*	95% CI [Lower, Upper]
Medial Hamstring	Intercept: NOCOG_AN	0.59	0.07	7.99	0.000	[0.45, 0.74]
	Cognitive Load: COG vs NOCOG	0.02	0.05	0.45	0.655	[-0.07, 0.11]
	Anticipation: UN vs AN	0.09	0.05	1.84	0.066	[-0.01, 0.18]
	Interaction: (COG vs NOCOG) × (UN vs AN)	-0.09	0.06	-1.48	0.139	[-0.22, 0.03]
	Entrance Time	-1.61	0.90	-1.80	0.073	[-3.38, 0.15]

Note. β = unstandardized coefficient; SE = standard error; CI = 95% confidence interval. The intercept represents the predicted mean for the reference condition (NOCOG_AN) and is not interpreted for significance. Pre-initial contact EMG amplitudes were calculated as the mean activation within the 100 ms window after initial force plate contact for the dominant limb. Bold values indicate $p < .05$.

Table 36. Fixed-effect results from the linear mixed-effects model on vastus medialis activation amplitude 50 ms prior to initial contact.

Variable	Term	β	SE	*z*	*p*	95% CI [Lower, Upper]
	Intercept: NOCOG_AN	0.45	0.06	8.19	0.000	[0.35, 0.56]
	Cognitive Load: COG vs NOCOG	0.04	0.04	0.97	0.334	[-0.04, 0.11]
Vastus Medialis	Anticipation: UN vs AN	0.04	0.04	1.08	0.280	[-0.03, 0.12]
	Interaction: (COG vs NOCOG) × (UN vs AN)	0.04	0.05	0.83	0.405	[-0.06, 0.15]
	Entrance Time	-0.22	0.75	-0.30	0.766	[-1.70, 1.25]

Note. β = unstandardized coefficient; SE = standard error; CI = 95% confidence interval. The intercept represents the predicted mean for the reference condition (NOCOG_AN) and is not interpreted for significance. Pre-initial contact EMG amplitudes were calculated as the mean activation within the 50 ms window preceding initial force plate contact for the dominant limb. Bold values indicate $p < .05$.

Table 37. Fixed-effect results from the linear mixed-effects model on vastus medialis activation amplitude 100 ms post initial contact.

Variable	Term	β	SE	*z*	*p*	95% CI [Lower, Upper]
Vastus Medialis	Intercept: NOCOG_AN	1.57	0.24	6.52	0.000	[1.10, 2.04]
	Cognitive Load: COG vs NOCOG	-0.22	0.21	-1.04	0.297	[-0.63, 0.19]
	Anticipation: UN vs AN	-0.24	0.21	-1.11	0.266	[-0.66, 0.18]
	Interaction: (COG vs NOCOG) × (UN vs AN)	0.14	0.29	0.48	0.631	[-0.44, 0.72]
	Entrance Time	2.30	4.14	0.55	0.579	[-5.81, 10.41]

Note. β = unstandardized coefficient; SE = standard error; CI = 95% confidence interval. The intercept represents the predicted mean for the reference condition (NOCOG_AN) and is not interpreted for significance. Pre-initial contact EMG amplitudes were calculated as the mean activation within the 100 ms window after initial force plate contact for the dominant limb. Bold values indicate $p < .05$.

Table 38. Fixed-effect results from the linear mixed-effects model on quadricep-hamstring ratio activation amplitude 50 ms prior to initial contact.

Variable	Term	β	SE	*z*	*p*	95% CI [Lower, Upper]
Q/H Ratio	Intercept: NOCOG_AN	0.73	0.16	4.47	0.000	[0.41, 1.04]
	Cognitive Load: COG vs NOCOG	0.23	0.11	2.19	**0.028**	[0.02, 0.44]
	Anticipation: UN vs AN	0.16	0.11	1.45	0.147	[-0.06, 0.37]
	Interaction: (COG vs NOCOG) × (UN vs AN)	-0.14	0.15	-0.91	0.362	[-0.43, 0.16]
	Entrance Time	2.27	2.10	1.08	0.279	[-1.84, 6.38]

Note. β = unstandardized coefficient; SE = standard error; CI = 95% confidence interval. The intercept represents the predicted mean for the reference condition (NOCOG_AN) and is not interpreted for significance. Pre-initial contact EMG amplitudes were calculated as the mean activation within the 50 ms window preceding initial force plate contact for the dominant limb. Bold values indicate $p < .05$.

Table 39. Fixed-effect results from the linear mixed-effects model on quadricep-hamstring ratio activation amplitude 100 ms post initial contact.

Variable	Term	β	SE	*z*	*p*	95% CI [Lower, Upper]
Q/H Ratio	Intercept: NOCOG_AN	3.67	0.60	6.13	0.000	[2.50, 4.85]
	Cognitive Load: COG vs NOCOG	-1.07	0.49	-2.21	**0.027**	[-2.03, -0.12]
	Anticipation: UN vs AN	-0.75	0.50	-1.52	0.129	[-1.73, 0.22]
	Interaction: (COG vs NOCOG) × (UN vs AN)	0.76	0.68	1.11	0.267	[-0.58, 2.09]
	Entrance Time	15.89	9.59	1.66	0.098	[-2.91, 34.70]

Note. β = unstandardized coefficient; SE = standard error; CI = 95% confidence interval. The intercept represents the predicted mean for the reference condition (NOCOG_AN) and is not interpreted for significance. Pre-initial contact EMG amplitudes were calculated as the mean activation within the 100 ms window after initial force plate contact for the dominant limb. Bold values indicate $p < .05$.

Appendix G. Individual Digit Recal Accuracy Scores

Table 40. Digit recall accuracy (%) across single- and dual-task conditions. Baseline is the single-test digit recall condition. Anticipated and Unanticipated are the cognitive dual-task condition

	Baseline	Anticipated	Unanticipated
S01	100	100	100
S03	91.7	100	100
S04	100	100	100
S05	91.7	100	100
S06	100	100	100
S07	88.3	100	86.7
S08	88.3	90.0	84.0
S09	96.7	86.7	93.3
S10	98.3	60.0	66.7
S11	95.0	100	100
S12	100	100	100
S13	98.3	100	100
S14	93.3	100	100
S15	100	100	100
S16	90.0	100	86.7
S17	100	90.0	100
S18	100	86.7	100
Mean ± SD	96.0 ± 4.5	94.9 ± 10.3	95.1 ± 9.2

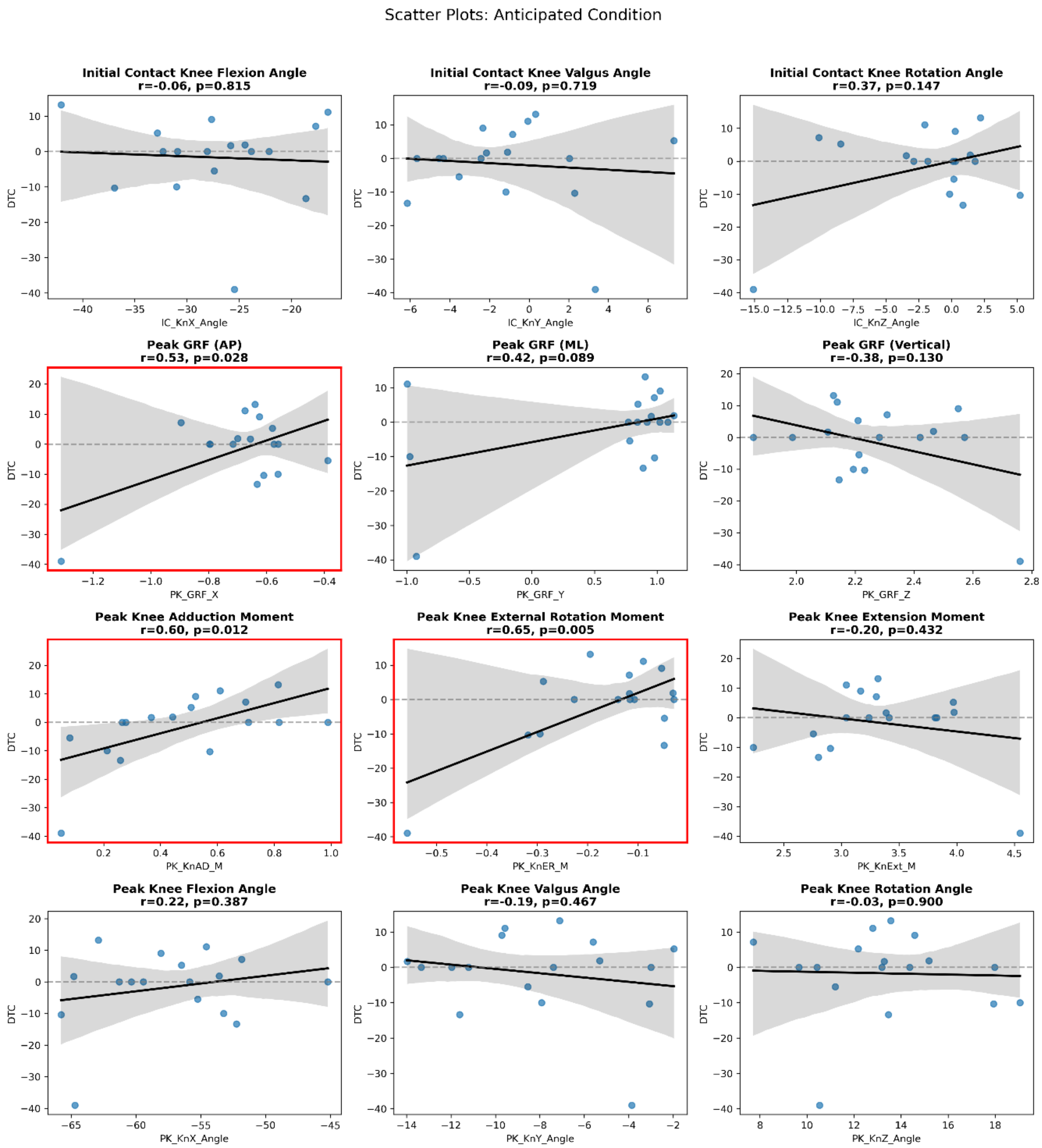

Figure 60. Scatterplots of the associations between dual-task cost (DTC) and knee joint mechanics and ground reaction forces during the anticipated cognitive condition. Scatterplots depict linear regression fits (black lines) with 95% confidence intervals (gray shading). Significant correlations ($p < 0.05$) are outlined in red.

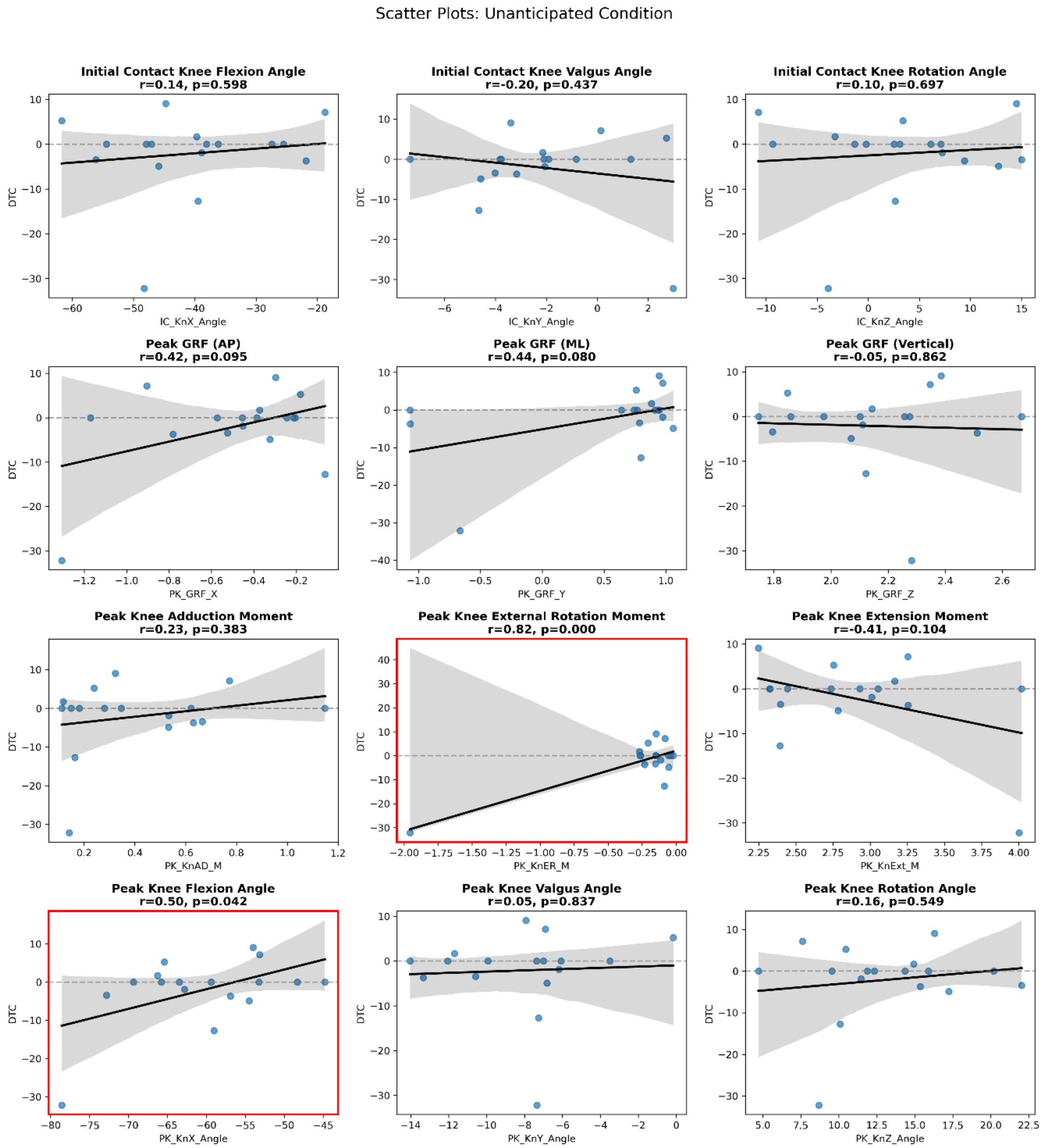

Figure 61. Scatterplots of the associations between dual-task cost (DTC) and knee joint mechanics and ground reaction forces during the unanticipated cognitive condition. Scatterplots depict linear regression fits (black lines) with 95% confidence intervals (gray shading). Significant correlations ($p < 0.05$) are outlined in red

Appendix I. Cognitive Motor Interference Classification Patterns

Table 41. Group-level cognitive-motor interference classification outcomes by number and percentage of participants (*n*) (%) during the dual-task conditions.

Initial Contact Knee Flexion Angle		
Framework Outcome	Anticipated	Unanticipated
Cognitive facilitation	7 (41.2)	2 (11.8)
Motor-related cognitive interference	5 (29.4)	4 (23.5)
Motor facilitation	--	3 (17.6)
Motor-priority trade-off	--	2 (11.8)
Mutual facilitation	--	2 (11.8)
No dual-task interference	5 (29.4)	4 (23.5)

Table 42. Group-level cognitive-motor interference classification outcomes by number and percentage of participants (n) (%) during the dual-task conditions.

Initial Contact Knee External/Internal Rotation Angle		
Framework Outcome	Anticipated	Unanticipated
Cognitive facilitation	7 (41.2)	4 (23.5)
Motor-related cognitive interference	5 (29.4)	5 (29.4)
Mutual interference	--	1 (5.8)
No dual-task interference	5 (29.4)	7 (41.2)

Table 43. Group-level cognitive-motor interference classification outcomes by number and percentage of participants (n) (%) during the dual-task conditions.

Initial Contact Knee Abduction/Adduction Rotation Angle		
Framework Outcome	Anticipated	Unanticipated
Cognitive facilitation	7 (41.2)	4 (23.5)
Motor-related cognitive interference	5 (29.4)	6 (35.3)
No dual-task interference	5 (29.4)	7 (41.2)

Table 44. Group-level cognitive-motor interference classification outcomes by number and percentage of participants (n) (%) during the dual-task conditions.

Peak Anteroposterior GRF		
Framework Outcome	Anticipated	Unanticipated
Cognitive facilitation	7 (41.2)	4 (23.5)
Motor-related cognitive interference	5 (29.4)	6 (35.3)
No dual-task interference	5 (29.4)	7 (41.2)

Table 45. Group-level cognitive-motor interference classification outcomes by number and percentage of participants (n) (%) during the dual-task conditions.

Peak Mediolateral GRF		
Framework Outcome	Anticipated	Unanticipated
Cognitive facilitation	7 (41.2)	2 (11.8)
Motor-related cognitive interference	5 (29.4)	4 (23.5)
Cognitive-related motor interference	--	1 (5.9)
Motor facilitation	--	2 (11.8)
Motor-priority trade-off	--	2 (11.8)
Mutual facilitation	--	2 (11.8)
No dual-task interference	5 (29.4)	4 (23.5)

Table 46. Group-level cognitive-motor interference classification outcomes by number and percentage of participants (n) (%) during the dual-task conditions.

Peak Vertical GRF		
Framework Outcome	Anticipated	Unanticipated
Cognitive facilitation	6 (35.3)	3 (17.6)
Motor-related cognitive interference	5 (29.4)	5 (29.4)
Mutual facilitation	1 (5.9)	1 (5.9)
Motor facilitation	--	2 (11.8)
Motor-priority trade-off	--	2 (11.8)
No dual-task interference	5 (29.4)	6 (35.3)

Table 47. Group-level cognitive-motor interference classification outcomes by number and percentage of participants (n) (%) during the dual-task conditions.

Peak Knee Adduction Moment		
Framework Outcome	Anticipated	Unanticipated
Cognitive facilitation	7 (41.2)	4 (23.5)
Motor-related cognitive interference	5 (29.4)	6 (35.3)
Motor facilitation	--	1 (5.8)
No dual-task interference	5 (29.4)	6 (35.3)

Table 48. Group-level cognitive-motor interference classification outcomes by number and percentage of participants (n) (%) during the dual-task conditions.

Peak Knee Extension Moment		
Framework Outcome	Anticipated	Unanticipated
Cognitive facilitation	7 (41.2)	3 (17.6)
Motor-related cognitive interference	5 (29.4)	5 (29.4)
Mutual interference	--	1 (5.9)
Cognitive-priority trade-off	--	1 (5.9)
No dual-task interference	5 (29.4)	7 (41.2)

Table 49. Group-level cognitive-motor interference classification outcomes by number and percentage of participants (n) (%) during the dual-task conditions.

Peak Knee External/Internal Rotation Moment		
Framework Outcome	Anticipated	Unanticipated
Cognitive facilitation	7 (41.2)	4 (23.5)
Motor-related cognitive interference	5 (29.4)	5 (29.4)
Motor-priority trade-off	--	1 (5.9)
No dual-task interference	5 (29.4)	7 (41.2)

Table 50. Group-level cognitive-motor interference classification outcomes by number and percentage of participants (n) (%) during the dual-task conditions.

Peak Knee Flexion/Extension Angle		
Framework Outcome	Anticipated	Unanticipated
Cognitive facilitation	7 (41.2)	4 (23.5)
Motor facilitation	--	1 (5.9)
Motor-related cognitive interference	5 (29.4)	5 (29.4)
Motor-priority trade-off	--	1 (5.9)
No dual-task interference	5 (29.4)	6 (35.3)

Table 51. Group-level cognitive-motor interference classification outcomes by number and percentage of participants (n) (%) during the dual-task conditions.

Peak Knee External/Internal Rotation Angle		
Framework Outcome	Anticipated	Unanticipated
Cognitive facilitation	7 (41.2)	4 (23.5)
Motor-related cognitive interference	5 (29.4)	6 (35.3)
No dual-task interference	5 (29.4)	7 (41.2)

Table 52. Group-level cognitive-motor interference classification outcomes. Group-level cognitive-motor interference classification outcomes by number and percentage of participants (n) (%) during the dual-task conditions.

Peak Knee Abduction/Adduction Angle		
Framework Outcome	Anticipated	Unanticipated
Cognitive facilitation	7 (41.2)	4 (23.5)
Motor-related cognitive interference	5 (29.4)	6 (35.3)
No dual-task interference	5 (29.4)	7 (41.2)

Appendix J. IRB Approval Letter

THE UNIVERSITY OF TENNESSEE KNOXVILLE

March 03, 2025

Joshua Trueblood Weinhandl, PhD
UTK - Coll of Education, Hlth, & Human - Kinesiology, Recreation & Sports Studies

Re: UTK IRB-25-08687-XP
Study Title: Cognitive-Motor Dual Tasks and Lower Extremity Mechanics

Dear Joshua Trueblood Weinhandl:

The UTK Institutional Review Board (IRB) reviewed your application for the above referenced project. It determined that your application is eligible for expedited review under 45 CFR 46.110(b)(1), categories 4, 6, and 7. The IRB has reviewed these materials and determined that they do comply with proper consideration for the rights and welfare of human subjects and the regulatory requirements for the protection of human subjects.

Therefore, this letter constitutes full approval by the IRB of your application (Version 1.3) as submitted, including Documents Stamped:

- Consent Form Revised - Version 4.0
- IRB_InstructionsScript_Estler - Version 1.0
- RecruitmentEmail _Estler - Version 1.0
- RecruitmentFlyer_Estler - Version 1.0
- Musculoskeletal Questionnaire_v2 - Version 1.0
- Fitness_Activity_Questionnaire - Version 1.0

that have been dated and stamped IRB approved. You are approved to enroll a maximum of 100 participants. Approval of this study will be valid from 03/03/2025 to 03/02/2028.

Any revisions in the approved application, consent forms, instruments, recruitment materials, etc., must also be submitted to and approved by the IRB prior to implementation. In addition, you are responsible for reporting any unanticipated serious adverse events or other problems involving risks to subjects or others in the manner required by the local IRB policy.

Finally, re-approval of your project is required by the IRB in accord with the conditions specified above. You may not continue the research study beyond the time or other limits specified unless you obtain prior written approval of the IRB.

Sincerely,

L Beebe

Lora Beebe, Ph.D., PMHNP-BC, FAAN

Institutional Review Board | Office of Research & Engagement
1534 White Avenue Knoxville, TN 37996-1529
865-974-7697 865-974-7400 fax irb.utk.edu

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Appendix K. Informed Consent Form

Consent for Research Participation

Research Study Title: Cognitive-Motor Dual Tasks and Lower Extremity Mechanics

Researcher(s): Joshua T. Weinhandl, PhD, University of Tennessee, Knoxville

Kaileigh E. Estler, M.S. University of Tennessee, Knoxville

Why am I being asked to be in this research study?

We are inviting you to participate in this study because you are a female between 18 and 30 years old, healthy, and engage in recreational activity at least three times a week for at least 30 minutes per session (e.g., running, weight training, elliptical, pick-up basketball, etc.). If you do not meet these criteria, have injured your legs in the past six months, or have had leg surgery, you will not be eligible for the study.

How long will I be in the research study?

If you agree to be in the study, your participation will involve 1 study visit lasting approximately 2 hours.

What will happen if I say "Yes, I want to be in this research study"?

If you agree to be in this study, we will ask you to come to the Biomechanics Lab for one testing session, lasting approximately 2 hours

Testing Preparation:

1. You will arrive wearing compression shorts, and a sports bra or athletic top. If you do not have these, they will be provided. Standardized lab shoes will be provided for you to change into.
2. The researcher will review the completed forms with you and provide an overview of the protocol.
3. Your height will be measured.
4. You will complete a cognitive digit recall test.
5. We will prepare/clean the skin so we can place sensors that measure muscle activity during the movement tasks. These sensors will be taped to the front and back of both thighs. This process will involve shaving, lightly scraping the skin, and cleaning it with alcohol where the sensors will be applied.
6. Next, you will complete a 3-minute warm-up jog at a self-selected pace.
7. You will complete trials of muscle contractions at your maximum effort.
8. Next, retroreflective markers will be placed on your trunk, pelvis, legs, and feet to create a representative model for motion analysis.
9. You will be given a heart rate monitor to wear on your arm during the run and cut. This monitor will track your heart rate to help assess your physical exertion levels. Between trials, you will be asked how you are feeling, and your heart rate will be monitored. You will be allowed to take sufficient rest between trials, based on your own needs.

Testing Procedures:

1. You will perform run and cut trials under four conditions: 1) unanticipated with no cognitive task, 2) anticipated with no cognitive task, 3) unanticipated with a cognitive task, and 4) anticipated with a cognitive task.
   - The direction of your cut will be determined by an arrow shown to you on the screen.
   - For anticipated trials, you will see the arrow before you begin running.

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IRB APPROVAL DATE: 03/03/2025
IRB EXPIRATION DATE: 03/02/2028

- For unanticipated trials, you will be shown the arrow while you run.
- After each trial with a cognitive task, you will be asked to recall a set of digits.

What happens if I say “No, I do not want to be in this research study”?

Participation in this study is entirely voluntary. You can choose not to participate now or withdraw later without any impact on your grades, your relationship with your instructors, or your standing with the University of Tennessee, Knoxville.

What happens if I say “Yes” but change my mind later?

You can choose to stop participating at any time, even if you initially agree to join. If you wish to withdraw, simply inform the PI and/or co-PI. Your information will remain de-identified and stored securely in a locked drawer in the Biomechanics lab. Only study personnel will have access to the data and forms collected prior to
withdrawal.

Are there any possible risks to me?

It is possible that someone could find out you were in this study or see your study information, but we believe this risk is small because of the procedures we use to protect your information. These procedures are described later in this form.

Possible risks include lower extremity injury and cardiovascular exertion during the running and cutting tasks. These risks will be minimized. The movements included are movements you should be familiar with, due to your being recreationally active. **In the *very* unlikely case that you hurt yourself during the study visit, emergency services will be contacted immediately for assistance. If you realize afterwards that you have become injured, you should seek medical assistance.**

Are there any benefits to being in this research study?

We do not expect you to benefit from being in this study. By participating in this study, you can help us learn more about the relationship between increasingly complex neurocognitive tasks and biomechanical precursors for ACL injuries. We hope the knowledge gained from this study will benefit others in the future.

Who can see or use the information collected for this research study?

We will protect the confidentiality of your information by keeping all forms in a locked drawer or on a password-encrypted computer drive in the Biomechanics lab, which is locked every day. Only the investigators conducting this study will have access to your personal information. If information from this study is published or presented at scientific meetings, your name and other personal information will not be used. We will make every effort to prevent anyone who is not on the research team from knowing that you gave us information or what information came from you. Although it is unlikely, there are times when others may need to see the information, we collect about you. These include people at the University of Tennessee, Knoxville who oversee research to make sure it is conducted properly.

Government agencies (such as the Office for Human Research Protections in the U.S. Department of Health and Human Services), and others responsible for watching over the safety, effectiveness, and conduct of the research. If a law or court requires us to share the information, we would have to follow that law or final court ruling.

What will happen to my information after this study is over?

We will keep your information to use for future research. Your name and other information that can directly identify you will be kept secure and stored separately from your research data collected as part of the study. We will not share your research data with other researchers.

Will I be paid for being in this research study?

You will not be paid for being in this study.

Will it cost me anything to be in this research study?

It will not cost you anything to be in this study.

What else do I need to know?

We may need to stop your participation in the study without your consent if you do not follow the study instructions, no longer meet the study's eligibility requirements, if your safety comes into question, or if the study is stopped for any reason.

The University of Tennessee does not automatically pay for medical claims or give other compensation for injuries or other problems should you realize you are injured outside of the study visit.

Who can answer my questions about this research study?

If you have questions or concerns about this study, or have experienced a research related problem or injury, contact the researchers, Dr. Joshua Weinhandl (**jweinhan@utk.edu**, 865-974-9556) or Kaileigh Estler (**kestler1@vols.utk.edu**, 847-624-9744).

For questions or concerns about your rights or to speak with someone other than the research team about the study, please contact:

Institutional Review Board
The University of Tennessee, Knoxville
Phone: 865-974-7697
Email: **utkirb@utk.edu**

IRB NUMBER: UTK IRB-25-08687-XP
IRB APPROVAL DATE: 03/03/2025
IRB EXPIRATION DATE: 03/02/2028

STATEMENT OF CONSENT

I have read this form, and the research study has been explained to me. I have been given the chance to ask questions and my questions have been answered. If I have more questions, I have been told who to contact. By signing this document, I am agreeing to be in this study. I will receive a copy of this document after I sign it.

Name of Adult Participant	Signature of Adult Participant	Date

Researcher Signature (to be completed at time of informed consent)

I have explained the study to the participant and answered all his/her questions. I believe that he/she understands the information described in this consent form and freely consents to be in the study.

Name of Research Team Member	Signature of Research Team Member	Date

IRB NUMBER: UTK IRB-25-08687-XP
IRB APPROVAL DATE: 03/03/2025
IRB EXPIRATION DATE: 03/02/2028

SUBJECT: ________ DATE: _________________

MUSCULOSKELETAL QUESTIONNAIRE

Spine / Low Back / Sacroiliac Joint:

Have You Ever Suffered An Injury To Your Spine / Low Back / Sacroiliac Joint?

☐ YES ☐ NO

- Please describe: ____
  ____
- Indicate when the injury occurred (e.g., date or time frame): ___________________________
  ____

Have You Ever Been Hospitalized For A Spine / Low Back / Sacroiliac Joint Injury?

☐ YES ☐ NO

- When? ____
- Please Describe ____
  ____

Have You Ever Had Surgery of Any Kind on Your Spine / Low Back / Sacroiliac Joint?

☐ YES ☐ NO

- When? ____
- Please Describe ____
  ____

Have You Ever Been Advised Not To Participate In Athletic Activities Due To A Spine, Low Back, and/or Sacroiliac Joint Injury?

☐ YES ☐ NO

- Please Describe ____
  ____

IRB NUMBER: UTK IRB-25-08687-XP
IRB APPROVAL DATE: 03/03/2025

SUBJECT: ________ DATE: _________________

Thigh / Hamstring / Quadriceps:

Have You Ever Suffered An Injury To Your Thigh, Hamstring, and/or Quadriceps?

☐ YES ☐ NO

- Please describe: ____
  ____
- Indicate when the injury occurred (e.g., date or time frame): ____________________________
  ____

Have You Ever Had Surgery For A Thigh, Hamstring, and/or Quadriceps Injury?

☐ YES ☐ NO

- When? ____
- Please Describe ____
  ____

Have You Ever Been Advised Not To Participate In Athletic Activities Due To A Thigh, Hamstring, or Quadriceps Injury?

☐ YES ☐ NO

- Please Describe ____
  ____

SUBJECT: ________   DATE: ________________

Hip/Groin

Have You Ever Suffered An Injury To Your Hip / Groin *(including hernias and/or sports hernias)?*

☐ YES ☐ NO

- Please describe: ____

  ____

- Indicate when the injury occurred (e.g., date or time frame): ______________________

  ____

Have You Ever Had Surgery For A Hip / Groin Injury?

☐ YES ☐ NO

- When? ____
- Please Describe ____

  ____

Have You Ever Been Advised Not To Participate In Athletic Activities Due To A Hip and/or Groin Injury?

☐ YES ☐ NO

- Please Describe ____

  ____

SUBJECT: ________ DATE: ________________

Knee / Patella:

Have You Ever Suffered An Injury To Your Knee and/or Patella (kneecap)?

☐ YES ☐ NO

- Please describe: ____
  ____
- Indicate when the injury occurred (e.g., date or time frame): ____________________________
  ____

Have You Ever Been Hospitalized For A Knee and/or Patella Injury?

☐ YES ☐ NO

- When? ____
- Please Describe ____
  ____

Have You Ever Had Surgery For A Knee and/or Patella Injury?

☐ YES ☐ NO

- When? ____
- Please Describe ____
  ____

Have You Ever Been Advised Not To Participate In Athletic Activities Due To A Knee / Patella Injury?

☐ YES ☐ NO

- Please Describe ____
  ____

SUBJECT: ________   DATE: _________________

Foot / Toes:

Have You Ever Suffered An Injury To Your Foot / Toe(s)?

☐ YES ☐ NO

- Please describe: ____

  ____

- Indicate when the injury occurred (e.g., date or time frame): ______________________________

  ____

Have You Ever Had Surgery For A Foot / Toe Injury?

☐ YES ☐ NO

- When? ____
- Please Describe ____

  ____

Have You Ever Been Advised Not To Participate In Athletic Activities Due To An Foot and/or Toe Injury?

☐ YES ☐ NO

- Please Describe ____

  ____

SUBJECT: ________ DATE: _________________

Ankle / Lower Leg:

Have You Ever Suffered An Injury To Your Ankle / Lower Leg?

☐ YES ☐ NO

- Please describe: ____

  ____

- Indicate when the injury occurred (e.g., date or time frame): ____________________________

  ____

Have You Ever Been Hospitalized For An Ankle / Lower Leg Injury?

☐ YES ☐ NO

- When? ____
- Please Describe ____

  ____

Have You Ever Had Surgery For An Ankle / Lower Leg Injury?

☐ YES ☐ NO

- When? ____
- Please Describe ____

  ____

Have You Ever Been Advised Not To Participate In Athletic Activities Due To An Ankle / Lower Leg Injury?

☐ YES ☐ NO

- Please Describe ____

  ____

SUBJECT: ________ DATE: _________________

FITNESS ACTIVITY QUESTIONNAIRE
Please describe your current participation in the following types of exercise:

1. *Aerobic (aerobic classes, walking, jogging, stair climbing, hiking, cycling, etc.)*
   Frequency (# of days per week): ______________ days
   Duration (time spent per session): ______________ minutes
   Intensity (difficulty level): light moderate vigorous

2. *Anaerobic (weight training, sprinting, etc.)*
   Frequency (# of days per week): ______________ days
   Duration (time spent per session): ______________ minutes
   Intensity (difficulty level): light moderate vigorous

3. *Organized or Recreational sports*
   Type of sport(s): ____
   Frequency (# of days per week): ______________ days
   Duration (years participating): ______________

IRB NUMBER: UTK IRB-25-08687-XP
IRB APPROVAL DATE: 03/03/2025

Appendix M. Participant Recruitment Flyer

Multitasking – we all think we're good at it, but are we really?

Join our study to help investigate how it impacts performance!

The University of Tennessee Biomechanics/Sports Medicine Lab is conducting a research study to investigate how cognitive-motor dual tasks influence lower extremity mechanics during run-and-cut movements.

WHO:

You May Qualify If You

- A female between 18 and 30 years old
- Recreationally active
- Have no previous lower extremity surgery ever
- No lower extremity injuries in the past 6 months

WHAT:

What does participation involve?

- One testing session, approximately 2 hours
- Run-and-cut trials under both anticipated and unanticipated conditions.
- For trials with a cognitive task, participants will be asked to recall a set of digits after completion of the run-and-cut.

WHEN:

Let's get started!

- If you are willing to participate, please scan the QR code to complete a brief screening survey. A researcher will then follow up with

For more information contact:

Kaileigh Estler (kestler1@vols.utk.edu)

Dr. Weinhandl (jweinhan@vols.utk.edu)

IRB APPROVAL DATE: 03/03/2025

Appendix N. Qualtrics Screening Survey

THE UNIVERSITY OF TENNESSEE KNOXVILLE

Demographics

. Please enter your first and last name:

. Please provide your email address for contact purposes:

. What is your biological sex?

- ○ Male
- ○ Female
- ○ Prefer not to say

. Please enter your age

Fitness Activity

. Please describe your *current* participation in ***aerobic*** exercise (e.g., aerobic classes, walking, jogging, stair climbing, etc.)

Frequency (# of days per week):

Duration (minutes spent per session)

Describe the intensity as one of the following: light, moderate, or vigorous

Type(s) of activity:

. Please describe your *current* participation in ***anaerobic*** exercise (e.g., weight training, sprinting, etc.)

Frequency (# of days per week):

Duration (minutes spent per session)

Describe the intensity as one of the following: light, moderate, or vigorous

Type(s) of activity:

. Please describe your *current* participation in organized or recreational sports

Type of sport(s):

Frequency (# of days per week):

Duration (years participating):

. Please list any *previous* experience with organized or recreational sports (e.g., soccer, basketball, lacrosse, volleyball, etc.)

Type of sport(s):

When (e.g., age, years):

Level (e.g., recreational, competitive):

Spine / Low Back / Sacroiliac Joint

Q1. Have you ever suffered an injury to your spine / low back / sacroiliac joint?

○ No
○ Yes

Q2. Have you ever been hospitalized for a spine / low back / sacroiliac joint injury?

○ No
○ Yes

Q3. Have you ever had surgery of any kind on your spine / low back / sacroiliac joint?

○ No
○ Yes

Q4. If you answered 'YES' for questions 1-3, please provide a brief description. You may skip this question if it does not apply.

Describe the type of spine, low back, sacroiliac joint injury:

Indicate when the injury occurred (e.g., date or time frame):

Specify any surgery performed as a result of injury:

Indicate what physical activities were refrained from and for how long:

Hip / Groin

Q5. Have you ever suffered an injury to your hip/groin (including hernias and/or sports hernias)?

○ No
○ Yes

Q6. Have you ever had surgery for a hip or groin injury?

○ No
○ Yes

Q7. If you answered 'YES' for questions 5–6, please provide a brief description.
You may skip this question if it does not apply.

Describe the type of hip / groin injury:	
Indicate when the injury occurred (e.g., date or time frame):	
Specify any surgery performed as a result of the injury:	
Indicate what physical activities you were refrained from and for how long:	

Thigh / Hamstring / Quadriceps

Q8. Have you ever suffered an injury to your thigh, hamstring, and/or quadriceps?

○ No
○ Yes

Q9. Have you ever had surgery for a thigh, hamstring, and/or quadriceps injury?

○ No
○ Yes

Q10. If you answered 'YES' for questions 8–9, please provide a brief description.
You may skip this question if it does not apply.

Describe the type of thigh, quadricep, hamstring injury:	
Indicate when the injury occurred (e.g., date or time frame):	

Specify any surgery performed as a result of the injury:

Indicate what physical activities you were refrained from and for how long:

Knee / Patella

Q10. Have you ever suffered an injury to your knee and/or patella (knee cap)?

○ No
○ Yes

Q11. Have you ever been hospitalized for a knee and/or patella injury?

○ No
○ Yes

Q12. Have you ever had surgery for a knee and/or patella injury?

○ No
○ Yes

Q13. If you answered 'YES' for questions 10–12, please provide a brief description.
You may skip this question if it does not apply.

Describe the type of knee / patella injury and when it occurred:

Indicate when the injury occurred (e.g., date or time frame):

Specify any surgery performed as a result of the injury:

Indicate what physical activities you were refrained from and for how long:

Ankle / Lower Leg

Q14. Have you ever suffered an injury to your ankle / lower leg?

○ No
○ Yes

Q15. Have you ever been hospitalized for an ankle / lower leg injury?

○ No
○ Yes

Q16. Have you ever had surgery for an ankle / lower leg injury?

○ No
○ Yes

Q17. If you answered 'YES' for questions 14-16, please provide a brief description. You may skip this question if it does not apply.

Describe the type of ankle / lower leg injury:

Indicate when the injury occurred (e.g., date or time frame):

Specify any surgery performed as a result of injury:

Indicate what physical activities you were refrained from and for how long:

Foot / Toes

Q18. Have you ever suffered an injury to your foot / toe(s)?

○ No

○ Yes

Q19. Have you ever had surgery for a foot / toe injury?

○ No
○ Yes

Q20. If you answered 'YES' for questions 18-19, please provide a brief description. You may skip this question if it does not apply.

Describe the type of foot / toe(s) injury: ☐

Indicate when the injury occurred (e.g., date or time frame): ☐

Specify any surgery performed as a result of injury: ☐

Indicate what physical activities you were refrained from and for how long: ☐

Q38. Please select at least 3 time slots, each lasting 2 hours, that you would be available for testing.

	Monday	Tuesday	Wednesday	Thursday	Friday	Saturday	Sunday
8:00 AM	☐	☐	☐	☐	☐	☐	☐
9:00 AM	☐	☐	☐	☐	☐	☐	☐
10:00 AM	☐	☐	☐	☐	☐	☐	☐
11:00 AM	☐	☐	☐	☐	☐	☐	☐
12:00 AM	☐	☐	☐	☐	☐	☐	☐
1:00 PM	☐	☐	☐	☐	☐	☐	☐
2:00 PM	☐	☐	☐	☐	☐	☐	☐
3:00 PM	☐	☐	☐	☐	☐	☐	☐
4:00 PM	☐	☐	☐	☐	☐	☐	☐
5:00 PM	☐	☐	☐	☐	☐	☐	☐
6:00 PM	☐	☐	☐	☐	☐	☐	☐

Vita

Kaileigh Estler was born in New Jersey in 1997, and has been fortunate to be raised by her four loving parents, Jim, Leslie, John, and Jen. She is the youngest of four sisters, Aniela and Alex Fila, and Courtney Donnelley. After high school, she pursued higher education, curious whether college was truly meant for her. Ironically, during that time, Kaileigh discovered a sense of purpose and academic success for the first time, which shaped her passion for learning and ultimately led her to pursue graduate studies in biomechanics.

She earned her Bachelor of Science and Masters of Science in Exercise Science from Old Dominion University, Norfolk, VA. Next, Kaileigh attended the Univeristy of Tennessee, Knoxville, where she completed her Doctor of Philosophy in Kinesiology and Sport Studies with a concentration in Biomechanics in 2025. During her doctoral studies, she also earned a Master's degree in Statistics. Kaileigh is grateful for the journey that has shaped her throughout her academic and doctoral studies, and will carry forward not only the knowledge gained, but also the professional and personal growth that have shaped who she is today. She plans to pursue a career that integrates her biomechanics expertise, data analytics skills, and research design experience across diverse fields.