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Integrated Aerodynamic and Biomechanical Analysis of Paralympic Tandem Time-Trial Cycling

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Abstract

Paralympic tandem time-trial cycling represents a complex performance problem in which two athletes form a coupled aerodynamic and biomechanical system. Aerodynamic interaction between pilot and stoker and posture-dependent geometric constraints introduce challenges that cannot be addressed through single-rider optimisation alone. The main scientific question addressed in this thesis is how aerodynamic performance and biomechanical sustainability interact across different tandem configurations, and whether aerodynamically optimised postures introduce measurable biomechanical penalties.

A fully integrated experimental–numerical framework was implemented. Computational fluid dynamics (CFD) simulations were performed on detailed three dimensional reconstructions of real pilot–stoker configurations. Multiple tandem setups were analysed by systematically varying pilot posture, stoker posture, and inter-rider spacing. Aerodynamic performance was quantified through effective drag area (C_dA), cumulative drag distributions, and qualitative flow-field analysis.

Across the investigated configurations, global C_dA varied by 7%. The results demonstrated that tandem performance is dominated by nonlinear interaction effects and that aerodynamic improvements are inherently non-additive. Posture modifications that reduce drag in isolation may be offset by unfavourable wake coupling within the two-rider system.

Biomechanical assessment was conducted using full-body inertial motion capture, musculoskeletal modelling, and surface electromyography (EMG) acquired under controlled laboratory conditions. Joint kinematics, cycle-averaged muscle activation patterns, and fatigue-related indicators were evaluated across riding positions. Results indicate that lower-limb kinematics and neuromuscular coordination timing remain robust despite substantial upper-body posture changes. Rather than inducing a global increase in fatigue, aerodynamically aggressive positions redistribute neuromuscular demand among muscle groups, typically shifting load from distal to proximal musculature.

These findings suggest that aerodynamic optimisation in tandem cycling does not inherently compromise biomechanical feasibility. Perceived discomfort in aggressive posi-

tions may be more closely related to training specificity than to fundamental mechanical limitations. The study highlights the necessity of system-level optimisation strategies that simultaneously consider aerodynamic interaction, structural constraints, and neuromuscular adaptation. Future work should extend this framework through experimental validation, metabolic assessment, and dynamic aerodynamic modelling to further refine system-level optimisation strategies in elite Paralympic tandem cycling.

Keywords: Cycling; Paralympic; Tandem; Aerodynamics; Computational Fluid Dynamics; CFD; Biomechanics; Electromyography; EMG; Sports Engineering.

Abstract in lingua italiana

Le prove a cronometro nel tandem paralimpico rappresentano un problema prestazionale complesso in cui due atleti costituiscono un sistema accoppiato dal punto di vista aerodinamico e biomeccanico. L'interazione aerodinamica tra pilota e stoker e i vincoli geometrici dipendenti dalla postura introducono criticità che non possono essere affrontate mediante un'ottimizzazione concepita per il singolo ciclista. La principale questione scientifica affrontata in questa tesi riguarda il modo in cui la prestazione aerodinamica e la sostenibilità biomeccanica interagiscono nelle diverse configurazioni del tandem, e se posture ottimizzate dal punto di vista aerodinamico comportino penalizzazioni biomeccaniche misurabili.

È stato implementato un approccio pienamente integrato di tipo sperimentale-numerico. Simulazioni di fluidodinamica computazionale (CFD) sono state eseguite su ricostruzioni tridimensionali dettagliate di configurazioni reali pilota-stoker. Sono state analizzate diverse configurazioni del tandem variando sistematicamente la postura del pilota, la postura dello stoker e la distanza longitudinale tra i due atleti. La prestazione aerodinamica è stata quantificata attraverso l'area equivalente di resistenza aerodinamica (C_dA), distribuzioni cumulative di drag e analisi qualitativa del campo di moto.

Tra le configurazioni investigate, il C_dA globale ha mostrato variazioni dell'ordine del 7%, evidenziando come la prestazione del tandem sia dominata da effetti di interazione non lineari e come i miglioramenti aerodinamici siano intrinsecamente non additivi. Modifiche posturali che riducono il drag in condizioni isolate possono infatti essere compensate da effetti sfavorevoli di accoppiamento della scia all'interno del sistema a due ciclisti.

La valutazione biomeccanica è stata condotta mediante acquisizione del movimento con sensori inerziali full-body, modellazione muscoloscheletrica ed elettromiografia di superficie (EMG) in condizioni di laboratorio controllate. Sono stati analizzati la cinematica articolare, i pattern di attivazione muscolare mediati sul ciclo di pedalata e indicatori legati alla fatica in diverse posizioni di guida. I risultati indicano che la cinematica degli arti inferiori e la tempistica della coordinazione neuromuscolare rimangono robuste nonostante significative variazioni posturali a livello del tronco e degli arti superiori. Piuttosto

che determinare un incremento globale della fatica, le configurazioni aerodinamicamente più aggressive redistribuiscono la richiesta neuromuscolare tra i diversi gruppi muscolari, tipicamente spostando il carico da muscoli distali a muscoli prossimali.

Questi risultati suggeriscono che l'ottimizzazione aerodinamica nel tandem paralimpico non comprometta intrinsecamente la fattibilità biomeccanica. Il disagio percepito in posizioni più aggressive può essere maggiormente correlato alla specificità dell'allenamento piuttosto che a limitazioni meccaniche. Lo studio evidenzia la necessità di strategie di ottimizzazione che considerino simultaneamente interazione aerodinamica, vincoli strutturali e adattamento neuromuscolare.

Sviluppi futuri dovrebbero estendere questo approccio mediante validazione sperimentale, valutazione degli aspetti metabolici e modellazione aerodinamica dinamica, al fine di affinare ulteriormente le strategie di ottimizzazione nell'ambito del ciclismo su tandem paralimpico d'élite.

Parole chiave: Ciclismo; Paralimpico; Tandem; Aerodinamica; Dinamica dei Fluidi Computazionale; CFD; Biomeccanica; Elettromiografia; EMG; Ingegneria dello Sport.

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1 | Introduction

1.1. Background and Motivation

Cycling performance emerges from the interaction between human physiology, mechanical systems, and environmental forces. Among all cycling disciplines, time-trial racing represents a particularly controlled and analytically tractable scenario, in which performance depends primarily on the balance between mechanical power production and resistive forces. For this reason, time trials have become a central reference for performance optimisation studies in competitive cycling, especially in relation to rider positioning and equipment design.

While a substantial body of research has addressed aerodynamic and biomechanical optimisation in able-bodied, single-rider cycling, Paralympic cycling—and tandem para-cycling in particular—has received comparatively limited scientific attention. This is notable given the additional aerodynamic interaction effects introduced by the presence of two riders in close proximity, whose combined postures influence the surrounding flow field in ways that differ from solo cycling. As a result, tandem performance is particularly sensitive to rider positioning and inter-athlete spacing [20, 21].

Although existing tandem studies have provided valuable insights, aerodynamic and biomechanical aspects are most often investigated separately. CFD analyses typically focus on configuration-level drag quantification using representative geometries, while biomechanical investigations in para-cycling predominantly examine physiological or kinematic variables without explicitly relating posture-dependent mechanical changes to their aerodynamic implications.

Systematic integration of aerodynamic simulations with experimentally acquired kinematic and electromyographic data on identical rider configurations remains extremely limited. The absence of such coupled analyses—combining motion capture, CAD reconstruction, CFD evaluation, and musculoskeletal modelling within a consistent framework—constitutes a significant gap in current performance optimisation strategies for tandem cycling [19, 22].

The present thesis addresses this gap by implementing a fully coupled experimental–numerical framework. Motion capture, musculoskeletal modelling, CAD reconstruction, and CFD are not treated as independent investigations, but as interconnected components of a single analytical pipeline. This integration enables direct assessment of whether aerodynamically favourable configurations compromise biomechanical feasibility, and vice versa—an evaluation that cannot be achieved through single-domain studies alone.

In this sense, the innovation of the present work lies not merely in analysing tandem aerodynamics or biomechanics, but in demonstrating that tandem time-trial optimisation is inherently a multi-domain problem requiring simultaneous consideration of fluid mechanics, structural constraints, and neuromuscular adaptation.

In professional tandem cycling, positioning strategies are often less standardised than in able-bodied single-rider disciplines. The relative scarcity of dedicated research and the reduced regulatory specificity allow for more substantial positional modifications, particularly in inter-rider spacing and cockpit configuration. Such macroscopic geometric adjustments may induce meaningful changes not only in aerodynamic behaviour but also in pedalling mechanics.

Beyond scientific motivations, this gap has an important ethical and social dimension. While able-bodied professional teams benefit from extensive access to sport science infrastructure and interdisciplinary research support, Paralympic teams often operate with substantially smaller budgets and limited access to wind tunnels, motion capture laboratories, or computational resources. Producing robust, transferable scientific knowledge in this domain can therefore contribute not only to performance enhancement but also to greater fairness and inclusivity in high-performance sport.

For these reasons, the present thesis aims to deliver a comprehensive and experimentally grounded analysis of tandem time-trial positions. By integrating motion capture, musculoskeletal modelling, CAD reconstruction, and CFD, this work seeks to identify posture-dependent aerodynamic and biomechanical trade-offs, contributing both to the scientific literature and to practical optimisation strategies for Paralympic tandem cycling.

1.2. Paralympic Sport and Tandem Cycling: Social and Historical Context

The development of Paralympic sport is closely linked to broader social and cultural movements centred on disability rights, accessibility, and inclusion. Originating in mid-twentieth-century rehabilitation programmes, Paralympic sport has progressively evolved



Figure 1.1: Paralympic tandem at the start of a time-trial race. Credit: Davide Andreoli

into a global high-performance sporting movement, accompanied by the establishment of international competitions and classification systems designed to ensure fairness while promoting athletic excellence [17].

Within this framework, para-cycling occupies a distinctive position due to its technological complexity and the diversity of functional classes involved. Tandem cycling, introduced to enable athletes with visual impairments to compete at high speed, represents one of the most emblematic disciplines. It requires a sighted pilot and a visually impaired stoker to function as a single coordinated unit, relying on trust, communication, and shared control. Performance therefore emerges not from individual capability alone, but from interpersonal coordination and system-level interaction.

Despite its symbolic and cultural importance, tandem cycling faces persistent structural constraints. Compared to able-bodied disciplines, tandem teams frequently report reduced access to advanced equipment, long-term research collaborations, and institutional support. The limited commercial market for tandem-specific technology further constrains innovation and slows the development of dedicated solutions [37].

Regulatory aspects introduce additional complexity. While the Union Cycliste Internationale (UCI) provides detailed and prescriptive regulations for conventional cycling,

tandem-specific provisions remain comparatively less explicit. This ambiguity may affect innovation and inter-team variability in equipment and rider positioning.

Understanding this social and institutional context is essential for framing the motivations and potential impact of the present work, which seeks to address performance optimisation within the practical constraints faced by Paralympic tandem teams.

1.3. Scientific Relevance of Aerodynamics and Biomechanics in Cycling

Cycling performance is constrained by both external resistive forces and the internal capacity of the athlete to generate sustainable mechanical power. Aerodynamics defines the external conditions under which power must be produced, while biomechanics governs how effectively this power can be generated and sustained without excessive fatigue or injury risk. These two domains therefore represent complementary and interdependent dimensions of performance optimisation.

In tandem cycling, this interdependence is particularly pronounced. The pilot and stoker form a single mechanical and aerodynamic system, such that posture optimisation in one domain may impose constraints in the other. Improvements achieved through aggressive positioning strategies may compromise biomechanical feasibility, reducing power output or increasing fatigue, while biomechanically favourable configurations may introduce performance penalties elsewhere. Tandem cycling thus represents a compelling case for integrated aerodynamic–biomechanical analysis, where optimisation cannot be pursued independently in either domain [20, 28].

Beyond competitive outcomes, such integration informs equipment design, posture optimisation, and athlete well-being. For Paralympic athletes—who may face additional physiological or sensory constraints—evidence-based approaches to these interactions are particularly important. Developing robust methodologies for studying these coupled effects therefore contributes to both performance optimisation and the broader advancement of inclusive sports engineering.

1.4. Institutional and Regulatory Framework (UCI and IPC Rules)

Competitive cycling is governed by a detailed regulatory framework designed to ensure fairness, safety, and technical consistency across events. For Paralympic cycling, this

framework is shaped primarily by two institutions: the Union Cycliste Internationale (UCI) and the International Paralympic Committee (IPC). Together, they define equipment standards, athlete classification systems, event formats, and safety protocols.[34, 35]

However, while UCI regulations for single bicycles are extensive and highly prescriptive—specifying geometries, dimensions, position limits, and component usage (Figure 1.2)—the equivalent framework for tandem bicycles is comparatively sparse. This discrepancy stems partly from the smaller number of teams competing in tandem events and partly from historical path dependencies in regulation development. The result is an environment where several aspects of tandem configuration remain ambiguously defined.[35]

The UCI Para-cycling Regulations outline general principles for tandem bicycles, including the requirement for two riders, the positioning of pilot and stoker, and the prohibition of certain aerodynamic components. Yet the rules do not provide the same granularity found in regulations for individual time-trial bicycles. Definitions of allowable cockpit geometry, maximum reach, stoker-specific extensions, and tandem-specific saddle placement remain either broadly stated or absent. For example, Article 1.3.008 of the UCI *General Organisation of Cycling as a Sport* regulations defines permissible points of support for a rider, yet its applicability to the stoker in tandem time trials—who often grips the frame rather than a conventional handlebar—is not explicitly clarified. This illustrates a broader pattern of interpretative uncertainty within the current regulatory framework [34].

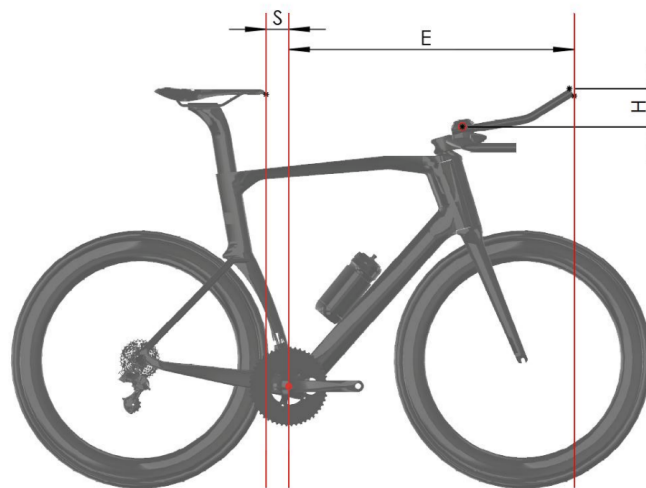


Figure 1.2: UCI regulations define geometric and contact-point constraints for able-bodied individual time-trial bicycles. In current practice, these constraints are generally applied to tandem pilots as well. For tandem stokers, however, the applicability of the same regulations is not explicitly defined and is often interpreted inconsistently, leading to heterogeneous positioning solutions across teams. Image source: UCI Regulations 2026.

The IPC contributes by defining athlete classification criteria, ensuring that visually impaired competitors meet standardized eligibility thresholds and compete in appropriate categories. Although these classifications ensure fairness from a physiological standpoint, they do not address the technical intricacies of tandem equipment.[35]

This regulatory landscape has several implications. Ambiguity can discourage innovation, as teams may avoid novel configurations for fear of post-hoc disqualification. It can also lead to substantial inter-team variation in positioning strategies and equipment choices, potentially influencing competitive balance. A clearer, tandem-specific regulatory framework would support both safety and performance, providing teams with precise guidelines while enabling responsible technological development.

1.5. Challenges in Research and Funding for Paralympic Sport

Despite the increasing visibility of the Paralympic Games, research across most Paralympic disciplines remains significantly underfunded relative to able-bodied sports. Para-cycling exemplifies this disparity. Advanced performance research—particularly in aerodynamics, biomechanics, and equipment optimisation—requires access to high-cost infrastructure such as wind tunnels, motion capture laboratories, additive manufacturing facilities, and high-performance computing clusters. The financial and logistical barriers to accessing these resources are substantial for many national Paralympic programmes.

The scarcity of funding produces several cascading challenges:

- **Limited sample sizes:** With fewer athletes and reduced opportunities for laboratory testing, studies often rely on single-case analyses.
- **Reduced technological access:** Many Paralympic teams operate without direct access to wind-tunnel time or advanced biomechanical tools.
- **Lower commercial interest:** The niche market for tandem equipment limits industry involvement and slows product innovation.
- **Geographical inequity:** Nations with strong Paralympic funding outperform others not only athletically but also scientifically.

Academic institutions play a critical role in addressing these limitations, often serving as the primary drivers of Paralympic research. Collaborations between universities, national sport federations, and Paralympic committees have enabled significant progress in recent

years, but overall output remains small compared to mainstream cycling research.

Because scientific understanding informs equipment design, training methodologies, and safety protocols, this disparity has real competitive and ethical consequences. Underdeveloped scientific support can exacerbate performance inequalities that stem not from athletic ability but from technological disadvantage. Consequently, advancing research in para-cycling—and particularly in tandem cycling—has both scientific and social motivation: it contributes to fairer competition, enhances athlete well-being, and promotes the broader mission of inclusion underpinning the Paralympic movement.[37]

1.6. Research Objectives and Thesis Structure

In light of the gaps identified in the previous sections, this thesis aims to provide a rigorous and integrated analysis of aerodynamic and biomechanical factors influencing performance in Paralympic tandem time-trial cycling. The study combines computational aerodynamics with experimental biomechanics, with the objective of evaluating how realistic posture variations affect both the aerodynamic resistance of the tandem system and the biomechanical demands imposed on the athletes. The analysis is grounded in data acquired from elite tandem cyclists and is designed to reflect practical configurations relevant to training and competition.

The specific objectives of this thesis are:

- **to quantify macroscopic posture-dependent differences in tandem aerodynamics** through computational fluid dynamics analyses of multiple realistic configurations;
- **to characterise biomechanical feasibility and neuromuscular response** using experimental kinematics and surface electromyography, supported by musculoskeletal modelling;
- **to identify trade-offs** between aerodynamic efficiency and biomechanical sustainability for both pilot and stoker;
- **to interpret the results in light of regulatory and practical constraints**, supporting evidence-based position optimisation;
- **to provide methodological and practical considerations** for future research and equipment development in Paralympic tandem cycling.

To address these objectives, the remainder of the thesis is organised as follows.

- **Chapter 2** reviews the state of the art in cycling aerodynamics and biomechanics, with particular emphasis on tandem and para-cycling research and on the experimental and numerical methodologies commonly adopted in this field.
- **Chapter 3** presents the aerodynamic methodology, describing the research design, posture reconstruction and CAD generation pipeline, computational domain definition, meshing strategy, numerical setup, and grid-independence assessment used in the CFD analyses.
- **Chapter 4** describes the biomechanical methodology, including the experimental protocols, motion capture and EMG acquisition, data processing, and the musculoskeletal modelling framework adopted for interpretation.
- **Chapter 5** reports the aerodynamic results of the CFD campaign, analysing global drag trends and posture-dependent flow features across the investigated tandem configurations.
- **Chapter 6** presents the biomechanical results, focusing on kinematic variables, EMG activation patterns, fatigue-related indicators, and their relationship with the analysed postures.
- **Chapter 7** integrates the aerodynamic and biomechanical findings in a unified discussion, highlighting performance trade-offs, practical implications for position selection, and limitations of the adopted approaches. Furthermore, it summarises the main conclusions of the thesis and outlines directions for future work.

Together, these chapters provide a comprehensive and experimentally grounded investigation of tandem time-trial performance, combining engineering analysis with biomechanical evidence to support informed optimisation strategies within the specific context of Paralympic sport.

2 | Literature Review

2.1. Aerodynamics in Competitive Cycling

Aerodynamics plays a central role in modern competitive cycling, particularly in disciplines such as time trials and track pursuit, where performance is determined by the delicate balance between power production and resistive forces. At speeds typical of elite racing—often above 45 km/h—the majority of the resistive load arises from aerodynamic drag, which generally accounts for between 80% and 90% of the total external resistance acting on the cyclist–bicycle system. This dominance stems from the quadratic dependency of drag on relative airflow velocity, which results in even modest changes in posture or equipment having measurable effects on speed.[8, 18]

The aerodynamic drag force F_D acting on a cyclist is usually described by the classical relation

$$F_D = \frac{1}{2}\rho v^2 C_d A,$$

where ρ is the air density, v is the cyclist’s relative velocity to the air, and $C_d A$ is the effective drag area. This parameter is of particular interest because it encapsulates both the pressure and viscous contributions to drag in a single quantity. Since the associated aerodynamic power requirement scales as

$$P_{\text{aero}} = F_D \cdot v = \frac{1}{2}\rho v^3 C_d A,$$

it becomes clear that a small reduction in $C_d A$ translates into a disproportionately large energetic advantage. A reduction of only 5% in $C_d A$, for example, can lead to time savings of several seconds over a standard 20 km time trial, demonstrating why aerodynamics is often described as “free speed” for competitive cyclists.[14]

The human body, however, is a bluff body, with limbs, clothing folds, and equipment features contributing to complex flow phenomena such as separation bubbles, shear layers,

and large-scale wakes. These flow structures depend sensitively on rider posture, helmet shape, arm configuration, shoulder width, and the alignment between the back and the helmet tail. Aerodynamic optimisation must therefore be understood not merely as a matter of reducing frontal area but as an attempt to control the behaviour of the entire flow field surrounding the cyclist.[8]

In tandem cycling, the complexity increases further. The presence of two riders in close succession produces a fundamentally different aerodynamic environment compared to single-rider configurations. The pilot generates the primary wake, while the stoker interacts with this wake, benefiting from reduced dynamic pressure but also contributing additional flow disturbances. This produces drag characteristics that cannot be approximated simply by doubling the drag of a single cyclist; instead, the total drag of a tandem configuration is typically 60–80% higher, depending on posture, spacing, and equipment. These differences emphasise the need for dedicated aerodynamic research rather than extrapolation from single-cyclist studies.[20, 21]

2.1.1. Historical Evolution of Aerodynamic Studies

The study of aerodynamics in cycling has evolved considerably over the past century. Early cyclists intuitively recognised the importance of airflow resistance, adopting lower and more compact postures long before formal aerodynamic analyses became available. The introduction of drop handlebars in the late nineteenth century reflects one of the earliest examples of aerodynamically motivated design, even though the underlying fluid dynamics were not yet scientifically understood.[8]

A major shift occurred in the mid-twentieth century with the rise of wind-tunnel research. Initially developed for aviation and automotive applications, wind tunnels provided a controlled environment in which the influence of posture and equipment on drag could be studied more rigorously. The 1970s and 1980s marked the first systematic use of wind tunnels for elite cycling, highlighted by Francesco Moser’s several hour-record attempts, which demonstrated the competitive impact of scientifically informed aerodynamic refinement.[8]

The subsequent decades saw exponential growth in the sophistication of aerodynamic studies. By the early 2000s, professional cycling teams began integrating aerodynamicists and engineers into their performance staff. At the same time, academic research expanded rapidly, aided by advances in experimental techniques such as particle-image velocimetry and high-precision force balances. This period also witnessed the emergence of computational fluid dynamics as a powerful complementary tool. Early CFD models were limited by computational resources and geometric simplifications, yet they laid the groundwork



Figure 2.1: Francesco Moser establishing a new hour record in 1988 on a purposefully built bicycle

for the high-resolution simulations used today.[18]

In the past 10–15 years, a new phase of aerodynamic research has emerged, characterised by the use of full-scale digital twins of athletes, 3D scanning technologies, and hybrid turbulence modelling approaches that combine Reynolds-averaged Navier–Stokes (RANS) equations with large-eddy simulation (LES) concepts. These tools have enabled a detailed understanding of flow separation, wake topology, and posture sensitivity. Researchers have been able to quantify how minor positional changes affect both local pressure distributions and overall drag, offering unprecedented insight into the physical mechanisms governing aerodynamic performance in cycling.[18]

This historical progression reflects a broader trend in sports science: the shift from intuition-driven optimisation to data-driven, experimentally validated aerodynamic design. Cycling has become one of the most technologically advanced sports in terms of aerodynamic research, though this sophistication is far more pronounced for single-rider disciplines than for tandem configurations.[37]

2.1.2. Methodologies: Wind Tunnel, Field Testing, and CFD

Modern aerodynamic research in cycling relies on a combination of wind-tunnel measurements, field testing, and computational fluid dynamics. Each methodology offers distinct



Figure 2.2: Remco Evenepoel undergoing wind tunnel tests at Politecnico di Milano. Credit: Team Soudal Quick-Step

advantages and limitations, and their integration provides the most comprehensive understanding of aerodynamic performance.[8]

Wind-tunnel testing remains the reference standard for controlled aerodynamic measurements. In these facilities, airflow speed, turbulence level, and yaw angle can be prescribed with high precision, allowing researchers to isolate the effects of posture or equipment. The use of multi-axis force balances enables direct measurement of drag and sometimes lift and side forces, with accuracies typically within 1–2%. Despite these strengths, wind tunnels impose constraints: athletes must hold a static position, pedalling motion is either absent or mechanically constrained, and support structures can influence the aerodynamic environment. For tandem bicycles, additional challenges arise due to the larger geometry and the need for custom mounting systems.[19]

Field testing offers complementary strengths by capturing aerodynamic behaviour under realistic riding conditions. Coast-down and virtual elevation methods estimate C_dA by analysing speed–power relationships, environmental conditions, and rolling resistance. Field tests allow athletes to ride naturally, with full pedalling dynamics, body movement, and environmental variability. However, these benefits come at the cost of reduced precision. Wind variability, surface imperfections, and measurement errors in power output can introduce uncertainty, making repeated trials and statistical filtering essential.[14]

Computational fluid dynamics has become a cornerstone of cycling aerodynamics due to its ability to visualise the complete flow field and to evaluate configurations that would be difficult or impractical to test experimentally. CFD simulations can reproduce spe-

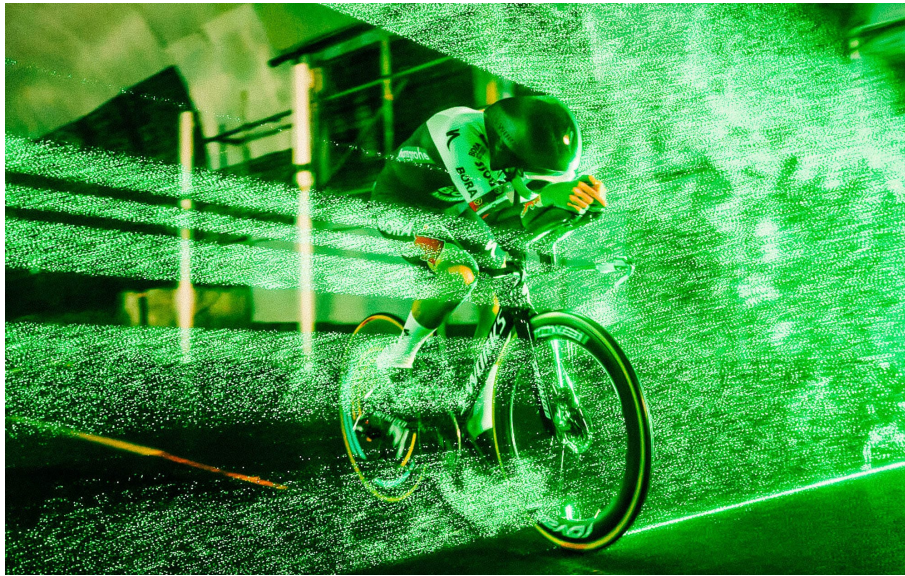


Figure 2.3: Particle image velocimetry is used with the purpose of validating CFD models. Credit: Team Red-Bull Bora

cific postures, examine the influence of equipment geometry, and explore environmental conditions such as crosswinds with precise control. Hybrid RANS–LES methods offer a balance between computational cost and accuracy, making them suitable for complex geometries such as cyclists and tandem bicycles. Typical mesh sizes for high-fidelity cycling simulations range from 20 to 80 million cells, depending on the level of anatomical detail and the domain size.[19]

Although CFD provides rich detail, it is inherently dependent on model assumptions, mesh quality, and turbulence modelling choices. For this reason, the most robust aerodynamic analyses integrate CFD results with wind-tunnel data and field measurements, using each methodology to validate the others. This triangulation approach has become standard practice in elite cycling research and is particularly important in tandem configurations, where wake interactions and posture sensitivities require high confidence in the underlying aerodynamic models.[8]

Together, these methodologies form a comprehensive framework for investigating cycling aerodynamics. Their combined use enables researchers to identify the most aerodynamically efficient positions while considering the practical constraints imposed by biomechanical feasibility, equipment regulations, and real-world riding conditions.

2.2. Biomechanics and Ergonomics in Cycling

While aerodynamic studies primarily focus on external flow phenomena and the resulting forces acting on the cyclist–bicycle system, biomechanical research addresses the internal constraints that determine whether aerodynamically favourable positions can be sustained efficiently over time and without increasing the risk of injury.

Cycling biomechanics examines how joint kinematics, muscular coordination, and body posture contribute to mechanical power production and to the long-term sustainability of effort. Unlike many athletic movements, cycling is characterised by a closed-chain and highly constrained motion: the pedals follow circular trajectories, the saddle constrains pelvic motion, and the handlebars largely define the position of the upper body. As a result, the relationship between posture, comfort, and performance emerges from a delicate balance between physiological capabilities and the geometric constraints imposed by the bicycle [14].

Understanding this interplay is essential in both able-bodied and Paralympic cycling contexts. Power production is not solely a function of maximal muscular capacity; it depends on the angles at which joints operate, the torque they can generate throughout the pedal cycle, and the neuromuscular strategies that distribute effort between different muscle groups. Similarly, comfort is not an accessory consideration but a determining factor in performance, particularly in time-trial events where athletes must maintain highly constrained postures for extended periods. Aerodynamic gains achieved through aggressive positions may be offset by reductions in power output or by the onset of discomfort that forces micro-adjustments, increasing metabolic cost or disturbing pedalling smoothness.[14]

In tandem cycling, these biomechanical considerations are further complicated by the presence of two riders. The pilot and the stoker inhabit different positions on the bicycle, experience different constraints, and assume distinct functional roles. The pilot is responsible for steering, balance, and pacing control, while the stoker generates power without direct control of the bicycle’s trajectory. This asymmetry produces unique mechanical and sensory conditions that influence posture, joint loading, and pedalling coordination. Moreover, the shared drivetrain enforces a common cadence, creating an additional layer of interdependence: both riders must adapt their individual pedalling styles to a unified rhythm, even if their natural cadence preferences differ.

2.2.1. Posture, Power Output, and Comfort

In cycling, posture is a determining factor in the expression of mechanical power. The ability of the lower limbs to generate torque depends critically on hip, knee, and ankle angles throughout the pedal cycle. The hip angle, in particular, governs the contribution of the gluteal and hamstring muscles. A more closed hip angle—defined by a reduced angle between the torso and the thigh—is typical of aerodynamic time-trial positions and has clear aerodynamic advantages. However, closing the hip beyond a certain threshold reduces the mechanical leverage of the gluteus maximus, forcing a redistribution of effort toward the quadriceps. This shift in muscular contribution can accelerate fatigue and alter the torque–angle relationship that characterises efficient pedalling.[2, 12, 32]

Knee kinematics play an equally important role. At bottom dead centre, the knee is typically flexed between 140° and 150° in efficient configurations. Larger angles may reduce muscular efficiency and increase stress on the patellar tendon, while smaller angles compromise mechanical advantage. Saddle height and fore–aft position must therefore be configured with millimetric precision to maintain an optimal balance between joint angles, force production, and comfort. Experimental studies have shown that deviations as small as 5 mm in saddle height can alter peak knee torque by several percent, subtly but meaningfully affecting long-term performance.[3, 15, 25]

Comfort influences these biomechanical mechanisms in ways that are both direct and indirect. A time-trial position that is aerodynamically favourable but induces discomfort in the cervical or lumbar spine may be unsustainable over race duration. Discomfort in the saddle area, typically caused by excessive pressure on soft tissues, can lead to frequent positional micro-adjustments, reducing pedalling smoothness and increasing metabolic cost. Similarly, a low-torso aerodynamic position can strain the neck as the athlete must tilt the head upward to maintain visibility, producing fatigue that spills over into the ability to stabilise the upper body.[24, 36]

Although such positions may be tolerable for the limited duration of a competition, their long-term sustainability during training is equally critical. If a posture restricts consistent force production or induces repeated mechanical overload on specific anatomical structures, it may limit the athlete’s ability to train effectively and could increase the risk of overuse injuries. Therefore, biomechanical feasibility should be evaluated not only in terms of acute race performance but also with respect to chronic adaptability and injury prevention.

In tandem cycling, these factors must be considered within a dual-rider system. The pilot often adopts a slightly more extended position to facilitate steering, while the stoker



Figure 2.4: Modern laser systems are nowadays used to measure the athlete's body angles, due to the importance of a correct bike fitting. Credit: Retul.

may be constrained by the geometry of the rear of the bicycle. The stoker's position frequently limits trunk rotation due to proximity to the pilot, which can influence pedalling smoothness and muscular recruitment. Additionally, the stoker's limited visual feedback affects proprioception and balance, often leading to a more rigid upper-body posture that, while stabilising, can increase muscular tension and reduce comfort. Postural adjustments must therefore be interpreted not only individually but also in terms of their influence on the joint rider system.[29, 31]

2.2.2. Existing Studies and Key Findings

Research on Paralympic tandem cycling has increased gradually over the past two decades, yet it remains considerably more limited than the extensive body of literature available on able-bodied cycling. The scientific contributions that do exist can be grouped into three loosely connected themes, although their boundaries are often blurred: performance studies based on competition data, aerodynamic investigations, and mechanical or equipment-focused analyses. Within each of these domains, researchers have attempted to quantify the distinctive characteristics of tandem performance, while acknowledging the scarcity of empirical data and the difficulty of generalising findings to a wider population of athletes.[37]

Retrospective performance analyses have provided some of the most informative long-term

perspectives. Studies examining results from Paralympic Games and major international competitions reveal a steady improvement in average speeds among tandem teams, often comparable in magnitude to the performance gains observed in elite able-bodied time trials. These trends are typically explained by advances in aerodynamics, equipment development, and training methods, but they also suggest increasing professionalism within Paralympic sport. At the same time, the distribution of performances across nations remains highly uneven, with countries investing more resources in Paralympic programmes consistently achieving better outcomes.[23]

From a more technical viewpoint, the most significant contributions arise from aerodynamic studies. Research on tandem aerodynamics has revealed that the system's behaviour differs fundamentally from that of single cyclists. The airflow around two riders seated in close succession is highly sensitive to torso angles, relative positioning, and the geometric relationship between the two bodies. Findings from wind-tunnel experiments and CFD analyses show that the wake generated by the pilot partially shelters the stoker, reducing their effective dynamic pressure but simultaneously exposing them to turbulent structures that are absent—or far less pronounced—in single-rider aerodynamics. Although this wake shielding can lower drag on the stoker, the combined drag of the system remains substantially higher than that of an individual cyclist, typically increasing by approximately 60–80%. This increase is not proportional to frontal area alone but arises from complex interactions between the two wakes, particularly at higher speeds or under crosswind conditions.[20, 22]

Studies focusing on torso-angle effects have shown that relatively small adjustments in pilot or stoker posture may produce measurable changes in drag. A reduction of a few degrees in the pilot's torso angle can yield reductions in C_dA of several percent, with practical time savings that are meaningful over typical time-trial distances. For the stoker, changes in posture produce smaller aerodynamic effects due to the pilot's wake, yet they still contribute to the overall flow field. These findings suggest that tandem posture optimisation cannot be approached as a simple extension of single-rider strategies; instead, it requires an understanding of the coupled aerodynamic system created by the two athletes.[21]

Crosswind behaviour has received particular attention because tandem bicycles present a larger lateral surface area than single bicycles. Under yawed inflow, the effective drag and side forces can increase significantly, influencing both speed and stability. Simulations show that yaw angles above 10° may induce substantial rolling moments, potentially affecting steering and control. These insights have implications not only for performance but also for safety, particularly in outdoor time trials where wind conditions can vary

unpredictably.[22]

Mechanical studies, although less numerous, contribute additional insight into the unique demands placed on tandem bicycles. The shared drivetrain transmits higher torque loads, and chain-drive efficiency can degrade under greater chain tension or misalignment. Frame stiffness requirements also differ, as the bicycle must accommodate the forces generated by two riders without compromising steering precision. These mechanical considerations are often overlooked in favour of aerodynamic factors, yet they play a non-negligible role in performance.[30]

Overall, the existing body of research has begun to reveal the distinctive physical characteristics of tandem cycling: the aerodynamics are more complex, the mechanical loads more substantial, and the performance determinants more intertwined than in single-athlete disciplines. Nevertheless, significant limitations remain, many of which stem directly from the methodological and practical challenges of studying a two-athlete system.[37]

3 | CFD Simulations Methodology

3.1. Overview of Research Design

The present study investigates the aerodynamic impact of different time-trial postures adopted by an elite Paralympic tandem cycling pair. The research design integrates anthropometric modelling, digital posture reconstruction, and numerical simulations based on the Reynolds-averaged Navier–Stokes (RANS) equations. The objective is to quantify how changes in pilot and stoker geometry influence the effective drag area C_dA under controlled and repeatable conditions. It should be emphasised that the aerodynamic analysis presented in this thesis represents a first-order, configuration-level investigation. The focus is placed on macro-positional changes (head alignment, torso inclination, inter-rider spacing, and cockpit configuration) rather than on fine-grained geometric details such as surface roughness, textile deformation, or minor anatomical adjustments. The objective is therefore to identify dominant aerodynamic mechanisms and interaction trends rather than to provide millimetric optimisation of local geometry.

The methodological workflow is structured into four main phases. First, the real competition posture was documented through high-resolution photographic acquisition, providing a basis for posture replication. Second, a detailed three-dimensional digital model of the athletes and tandem bicycle was reconstructed using Blender and MakeHuman, allowing precise manipulation of joint angles and inter-rider spacing. Third, a set of aerodynamic configurations (v0–v8) was generated, including both tandem and single-rider reference cases, each representing a realistic or strategically relevant variation of the time-trial setup. Finally, each configuration was simulated using OpenFOAM v2406 and post-processed in ParaView and MATLAB to extract aerodynamic forces and flow-field characteristics.

This chapter describes in detail the experimental setup, geometry-generation protocol, numerical methods, and the procedures used to ensure consistency, repeatability, and controlled comparison between configurations.



Figure 3.1: Picture of the athletes during a TT race, later used as a reference for 3D modeling. Credit: Davide Andreoli

3.2. Experimental Setup

3.2.1. Athletes and Equipment

Two elite Paralympic athletes served as subjects of this study (Figure 3.1). The stoker is 26 years old, with a height of 1.70 m and a mass of 65 kg, while the pilot is 28 years old, measuring 1.76 m and weighing 70 kg. Both athletes are current World Champions in road racing and aim to improve their performance in time-trial events, where aerodynamic optimisation plays a decisive role [23].

Their competition tandem bicycle is a high-performance model compliant with UCI para-cycling regulations [35]. The exact commercial specification is omitted for confidentiality, but its geometry was reconstructed directly from the manufacturer’s blueprint in Figure 3.2. The frame model includes accurate tube lengths, cockpit geometries, wheelbase, and wheel dimensions. The pilot uses a time-trial cockpit with aerobars, while the stoker’s cockpit varies depending on the configuration (frame clench, road handlebar, or time-trial extensions).

3.2.2. Data Acquisition Protocols

A two-stage data acquisition approach was initially considered. High-resolution photogrammetry was attempted using a professional camera and Polycam. Although the reconstructed point clouds captured global anthropometric proportions, the resulting meshes were too noisy, non-watertight, and unsuitable for CFD-grade surface discretisation [33].

A hybrid anthropometric reconstruction strategy was therefore adopted. MakeHuman



Figure 3.3: 3D modeling process of the current athletes' position (v0)



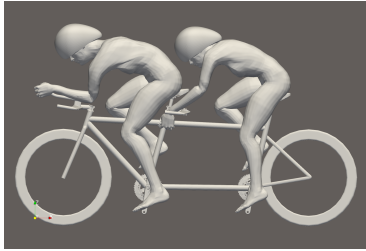
Figure 3.4: The Spanish Tandem team adopted a different position, with the stoker placing his hands on aero bars rather than gripping the frame. Credit: Davide Andreoli.

3.2.3. Time-Trial Positions Considered

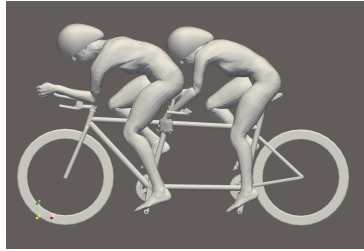
Nine configurations were analysed to investigate how posture, head angle, cockpit geometry, and stoker–pilot spacing influence aerodynamic drag:

- v0** Baseline race posture. Stoker in “frame clench” position, with a gap between the athletes’ backs. This is the most common configuration in the tandem time-trial racing scene: the stoker grips the frame directly, with a small handlebar functioning as a forearm support.
- v1** Stoker advanced by 7 cm to reduce the inter-rider separation, frame clench position, pilot unchanged with respect to v0
- v2** Pilot with lowered head and overall improved aerodynamic position, stoker unchanged with respect to v0.
- v3** Stoker moved 20 cm rearwards, equipped with time-trial extensions and head lowered, pilot unchanged with respect to v0.
- v4** Same inter rider spacing as v0, but the stoker has a standard road handlebar instead of gripping the frame, pilot unchanged with respect to v0.
- v5** Combination of v1 and v2: stoker advanced by 7 cm to reduce the inter-rider separation and pilot with lowered head and overall improved aerodynamic position
- v6** Combination of v2 and v3: stoker moved 20 cm rearwards, equipped with time-trial extensions and head lowered, and pilot with lowered head and overall improved aerodynamic position.
- v7** Pilot only in baseline position.
- v8** Pilot only in improved aerodynamic position.

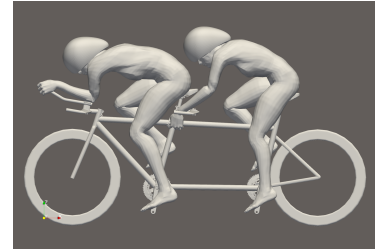
These models are shown in 3.5. These postures represent realistic adjustments that the athletes could adopt during training or competition. They allow the isolation of key factors such as torso angle, longitudinal spacing, and cockpit ergonomics. It is worth noting that many of these variations require a new frame to be built with different geometries; this study aims at analysing different strategies for the project of a future new time-trial tandem frame.



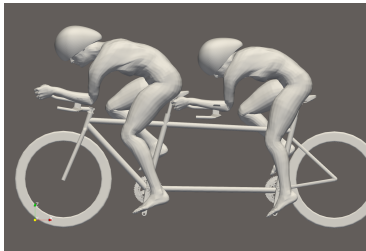
(a) v0 – Baseline tandem configuration



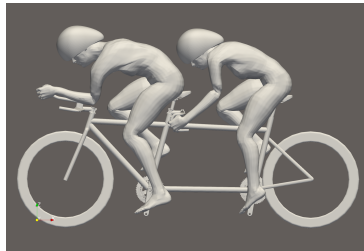
(b) v1 – Stoker advanced by +7 cm



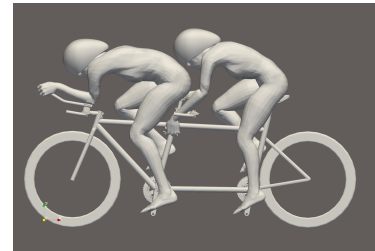
(c) v2 – Lowered pilot head position



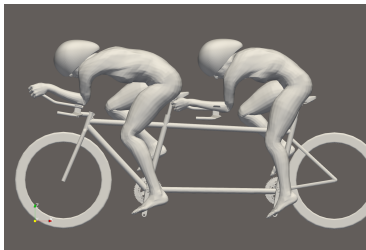
(d) v3 – Rearward stoker with TT extensions



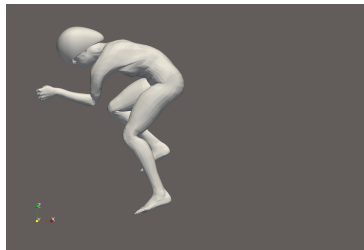
(e) v4 – Rearward stoker with road handlebar



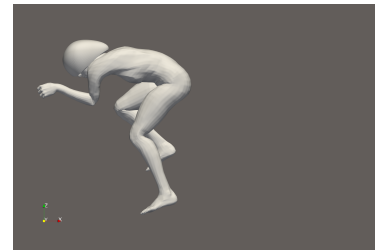
(f) v5 – Pilot lowered and stoker advanced



(g) v6 – Rearward stoker with TT extensions + lowered pilot head



(h) v7 – Pilot only, baseline position



(i) v8 – Pilot only, aerodynamic position

Figure 3.5: Three-dimensional CAD models of the nine configurations analysed in the CFD study (v0–v8).

3.3. CFD Analysis Framework

3.3.1. Geometry Definition and Computational Domain

All CFD simulations were performed using OpenFOAM v2406 adopting a full-domain approach. The computational domain was generated using the `blockMesh` utility and represents an idealised wind-tunnel-like environment designed for external aerodynamic analysis of cyclist configurations.

The choice of domain size was guided by established theoretical recommendations and previous numerical studies on cyclist and tandem aerodynamics [19, 33], which typically scale the computational domain with respect to a characteristic dimension of the model. In the present study, the characteristic height of the tandem system was estimated as:

$$H_{\text{model}} = 1.40 \text{ m.}$$

Accordingly, the domain height corresponds to approximately $5.7 H_{\text{model}}$, while the domain width was set equal to the height to minimise lateral confinement effects. The tandem geometry, with an overall length of approximately 2.4 m, was positioned such that the front wheel was located 6 m downstream of the inlet boundary, corresponding to approximately $4.3 H_{\text{model}}$.

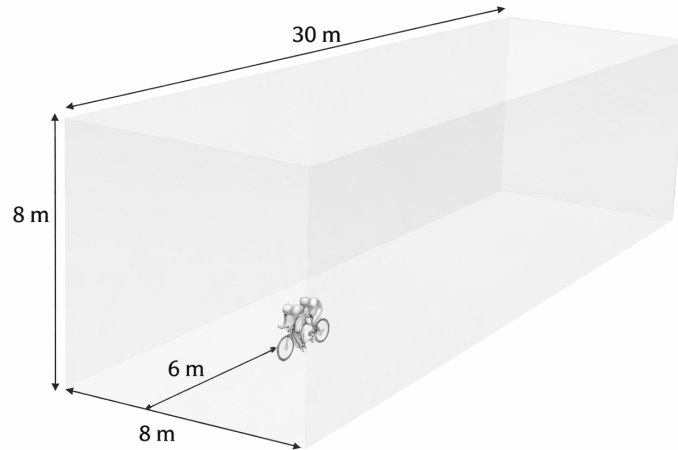


Figure 3.6: Domain dimensions

All simulations were performed using a full-domain setup, without imposing any symmetry plane. Although pedalling motion was not explicitly modelled, fixing the leg positions

while adopting a full-domain approach avoids artificial symmetry constraints and allows asymmetric wake structures to develop naturally.

The blockage ratio was evaluated to verify that confinement effects were negligible. The maximum frontal area of the tandem configuration was estimated as approximately 0.30 m^2 . Considering an inlet cross-sectional area of 64 m^2 , the resulting blockage ratio remains below 0.5%, well under the conventional 3% threshold [33]. The domain dimensions are summarised in Figure 3.6.

3.3.2. Sensitivity to Domain and Meshing Choices

Prior to the final simulation campaign presented in this work, several additional CFD simulations were conducted using alternative domain dimensions and preliminary surface-refinement strategies. Although these configurations initially yielded drag coefficients within ranges commonly reported in cycling aerodynamics literature, a more detailed analysis revealed inconsistencies that were not physically justifiable.

In particular, the cumulative drag distributions evaluated along the streamwise and vertical directions exhibited trends that could not be explained by geometric or postural variations. Local drag contributions appeared artificially amplified or attenuated in specific regions of the wake, and the relative ranking of configurations changed depending on domain size and refinement settings. Such behaviour indicated that the computed drag differences were partially influenced by numerical artefacts rather than solely by physical flow mechanisms.

These observations exposed a pronounced sensitivity of tandem simulations to domain definition and mesh resolution, especially in a multi-body configuration characterised by strong wake interaction and recirculation regions. In tandem cycling, the aerodynamic performance of the rear rider depends critically on the accurate resolution of the pilot's near-wake and base pressure recovery. Insufficient downstream domain length, inadequate lateral clearance, or overly coarse wake refinement can artificially constrain wake expansion or distort pressure recovery, leading to misleading drag predictions.

The issue is particularly relevant in interaction-driven systems, where small numerical distortions in the wake structure can propagate and significantly alter the drag of the downstream body. As highlighted in previous studies [4], rigorous domain verification is essential to avoid contamination of interaction effects by boundary-induced artefacts.

For these reasons, the domain setup ultimately adopted in the present study features extended downstream length and targeted wake refinement. The final configuration was

selected only after ensuring stability of global drag values and cumulative drag distributions with respect to further mesh refinement.

This sensitivity analysis constitutes a crucial methodological step of the present work and reinforces the necessity of systematic domain verification in multi-body aerodynamic simulations.

3.3.3. Mesh Generation Strategy

The computational mesh was generated using the `snappyHexMesh` utility, starting from a structured hexahedral background grid created with `blockMesh`. The meshing strategy was designed to ensure sufficient spatial resolution in regions of high aerodynamic relevance while maintaining computational feasibility for a large parametric study.

Local mesh refinement was applied through a hierarchy of seven volumetric refinement boxes positioned around the tandem geometry and extending downstream into the near-wake region. These refinement regions were defined to progressively increase spatial resolution from the far field towards the riders, with particular attention to the interaction region between pilot and stoker and to the wake development behind the rear athlete. The refinement regions are visible in Figure 3.7.

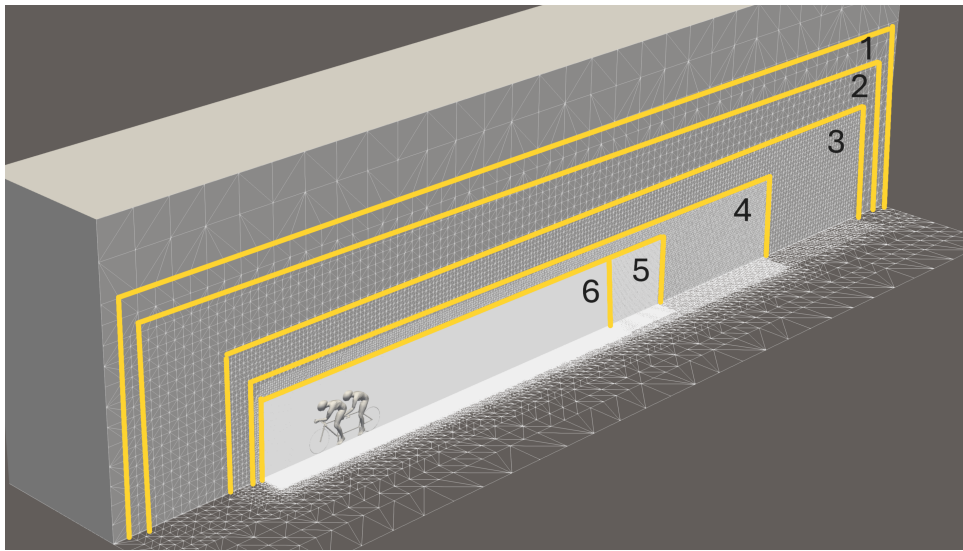
In addition to volumetric refinement, a distance-based refinement was applied in proximity to the tandem geometry to ensure adequate resolution of the near-body flow field. This refinement strategy was kept consistent across all meshes in terms of layout and spatial extent, while the refinement levels were systematically varied to construct meshes of increasing resolution.

Three different mesh resolutions were generated for mesh-independence assessment:

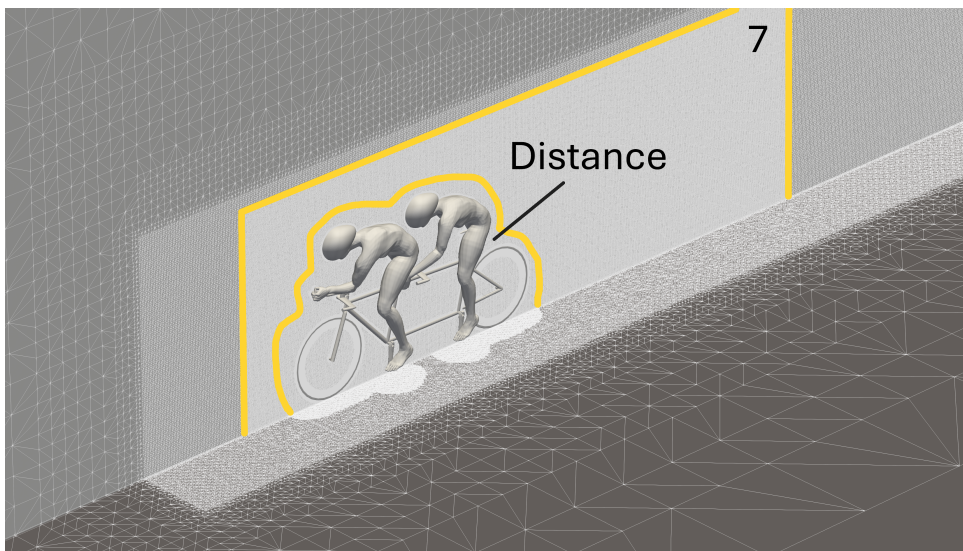
- a coarse mesh,
- a medium mesh,
- a fine mesh.

All three meshes share the same computational domain, geometry positioning, and refinement-box configuration. Differences between them arise exclusively from the refinement levels assigned to the volumetric regions, ensuring that any variation in aerodynamic results can be attributed solely to mesh resolution effects.

A comparative summary of the refinement levels adopted for the coarse, medium, and fine meshes is reported in Table 3.2. The quantitative mesh-independence analysis and final mesh selection are discussed in Section 3.3.6.



(a)



(b)

Figure 3.7: Refinement boxes inside the computational domain.

3.3.4. Near-Wall Treatment and Wall Resolution

The near-wall treatment represents a critical aspect in the CFD simulation of cycling aerodynamics, due to the complex geometry of the human body and the strong interaction between boundary layers, separation regions, and wake structures. In the present study, particular attention was dedicated to the assessment of wall resolution and the suitability of different near-wall modelling approaches.

Initial meshing attempts included the introduction of near-wall prism layers with the objective of resolving the viscous sublayer and directly capturing the boundary-layer velocity profile. Several configurations were tested by varying the first-layer thickness, growth ratio, and local refinement levels. However, despite extensive tuning, it was not possible to obtain a consistently low and uniform y^+ distribution across the entire tandem geometry. Localised regions of excessively high y^+ values systematically emerged, particularly in areas characterised by sharp curvature changes, close proximity between limbs and torso, and complex surface interactions between the two riders.

These difficulties are intrinsic to the aerodynamic simulation of human geometries and are further exacerbated in tandem configurations by the presence of multiple interacting bodies. As a consequence, a fully resolved boundary-layer approach could not be achieved in a robust and repeatable manner across all tested configurations. This limitation motivated the adoption of a wall-function-based near-wall treatment for the final simulation setup.

In the adopted meshing strategy, standard wall functions were employed in conjunction with the SST k - ω turbulence model. Specifically, the following wall-function formulations were used:

- `nutkWallFunction` for the turbulent kinematic viscosity,
- `kqRWallFunction` for the turbulent kinetic energy k ,
- `omegaWallFunction` for the specific dissipation rate ω .

The wall resolution was evaluated a posteriori by analysing the distribution of the non-dimensional wall distance y^+ on all relevant surfaces. For the medium-resolution mesh selected for production simulations, the average wall resolution over the cyclist and bicycle surfaces was found to lie in the range:

$$y_{\text{mean}}^+ \approx 25\text{--}35,$$

while the maximum observed values remained below:

$$y_{\max}^+ < 200.$$

These values are consistent with the recommended operating range for wall-function-based RANS simulations using the SST $k-\omega$ model, particularly for external aerodynamics applications involving complex and highly curved geometries. The absence of extensive regions with excessively high y^+ values ensures that the adopted wall functions operate within their intended validity range.

Given the comparative nature of the present study, which focuses on relative aerodynamic differences between tandem configurations rather than on absolute drag prediction, the selected near-wall treatment provides a reliable and computationally efficient solution. The consistency of wall-resolution characteristics across all simulated configurations further ensures that differences in aerodynamic performance are not influenced by variations in near-wall discretisation.

3.3.5. Boundary Conditions and Turbulence Specification

Boundary conditions were prescribed consistently across all CFD simulations to ensure direct comparability between the different tandem configurations. The flow was assumed incompressible and isothermal. At the inlet, a uniform freestream velocity was imposed with magnitude

$$U_{\infty} = 15 \text{ m/s},$$

representative of elite-level time-trial conditions. The outlet boundary was defined using a zero static gauge-pressure condition, while allowing the remaining transported variables to adjust smoothly through inlet–outlet type conditions.

All cyclist and bicycle surfaces were treated as no-slip walls. The ground boundary was modelled as a moving wall with imposed velocity equal to the freestream speed ($U = U_{\infty}$), in order to reproduce a rolling-road condition and avoid the formation of an artificial near-ground boundary layer. All remaining domain boundaries were treated as slip walls in order to approximate far-field conditions and minimise confinement effects.

The boundary conditions used in the simulations are summarised in Table 3.1.

Table 3.1: Boundary conditions

	Inlet	Outlet	Cyclists	Ground
P	zeroGradient	fixedValue	zeroGradient	zeroGradient
U	fixedValue	inletOutlet	noSlip	fixedValue
k	fixedValue	inletOutlet	kqRWallFunction	slip
ω	fixedValue	inletOutlet	omegaWallFunction	slip
ν_t	calculated	calculated	nutkWallFunction	calculated

Turbulence was modelled using the SST k - ω closure. Inlet turbulence quantities were prescribed following standard empirical relations based on turbulence intensity and a characteristic length scale. A turbulence intensity of

$$TI = 2\%$$

was assumed. The turbulent kinetic energy at the inlet was computed as

$$k = \frac{3}{2} (U_\infty TI)^2 = \frac{3}{2} (15 \times 0.02)^2 = 0.135 \text{ m}^2/\text{s}^2.$$

The specific dissipation rate was defined as

$$\omega = \frac{\sqrt{k}}{C_\mu^{1/4} L_{\text{ref}}}, \quad C_\mu = 0.09,$$

where the reference length scale was set equal to 10% of the tandem length:

$$L_{\text{ref}} = 0.1 L_{\text{tandem}} = 0.1 \times 2.4 = 0.24 \text{ m}.$$

This yields

$$\omega = \frac{\sqrt{0.135}}{0.09^{1/4} \times 0.24} \approx 2.8 \text{ s}^{-1}.$$

All boundary-condition definitions and turbulence inlet parameters were kept identical across all simulated cases to ensure that differences in aerodynamic forces arise exclusively from geometric variations associated with rider posture and inter-athlete spacing.

3.3.6. Mesh Independence and Grid Convergence Analysis

Due to the strong sensitivity of external aerodynamics simulations to mesh resolution, a dedicated grid independence study was performed before selecting the reference mesh for the full campaign. Three progressively refined meshes were generated by modifying

the refinement levels of the volumetric refinement boxes while keeping all other numerical settings unchanged.

The three meshes are referred to as:

- **Coarse mesh,**
- **Medium mesh,**
- **Fine mesh.**

The refinement strategy was identical in terms of box positioning and hierarchy, while the refinement levels assigned to each region were increased consistently from the coarse to the fine configuration. This approach ensured that grid convergence could be assessed without introducing geometric or numerical inconsistencies.

A comparative summary of the refinement levels applied to each refinement box is reported in Table 3.2. The total number of cells for each mesh and the corresponding aerodynamic force results are reported in Table 3.3.

Table 3.2: Refinement levels applied to the volumetric refinement boxes for the three mesh resolutions.

Refinement region	Coarse	Medium	Fine
RefinementBox1	2	2	3
RefinementBox2	3	3	4
RefinementBox3	4	4	5
RefinementBox4	5	5	6
RefinementBox5	5	6	7
RefinementBox6	6	6	7
RefinementBox7	6	7	7
Distance	7	7	7

Table 3.3: Grid independence study based on aerodynamic drag force.

	Coarse	Medium	Fine
Number of cells [–]	9.53×10^6	3.21×10^7	8.24×10^7
Mean drag force F_x [N]	37.67	36.51	36.89
Std. deviation (last 500 it.) [N]	0.49	0.57	0.38

All simulations were run for a total of 2500 iterations. Aerodynamic forces were monitored throughout the simulation, and statistical averaging was performed over the last 500 iterations, after force convergence had been reached. The force histories and residual evolution used to assess convergence are reported in Figures 3.8 and 3.9.

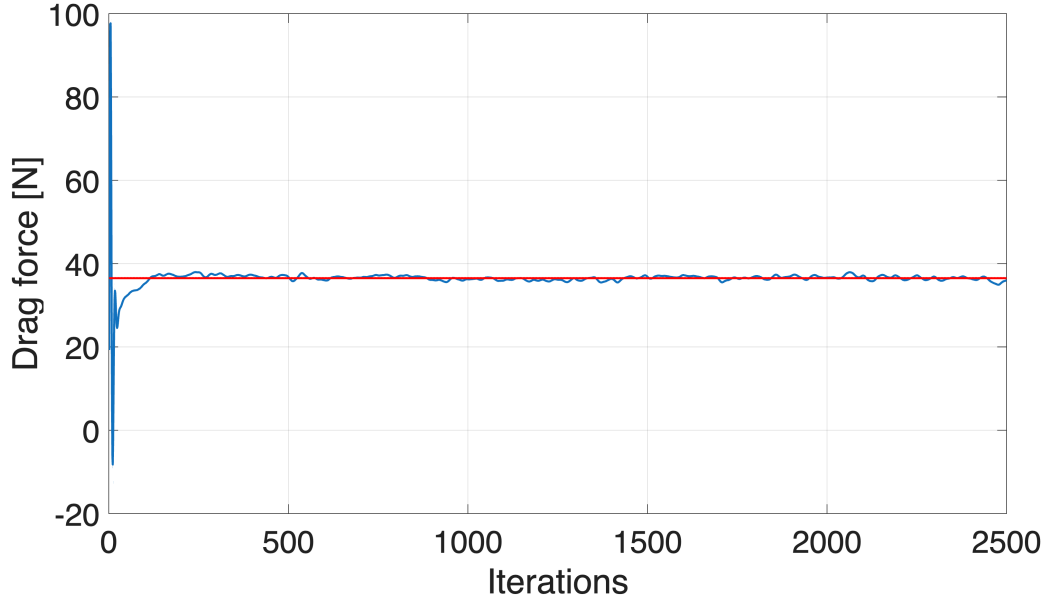


Figure 3.8: Drag force history for the medium mesh resolution.

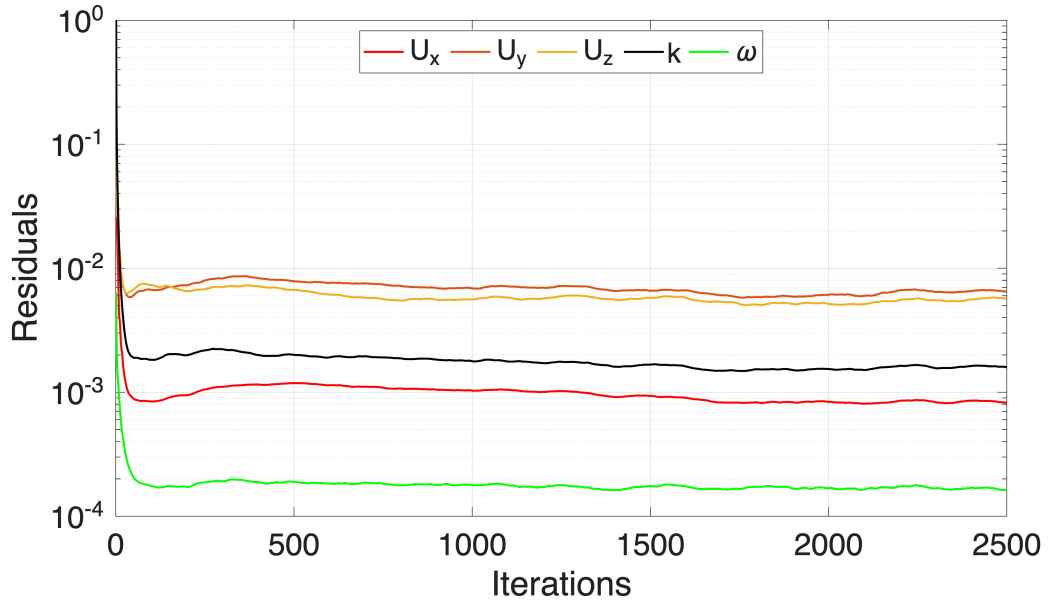


Figure 3.9: Residual convergence history for velocity and turbulence variables in the medium mesh resolution.

To quantify grid convergence and estimate the numerical uncertainty associated with mesh resolution, a Grid Convergence Index (GCI) analysis was performed following the standard procedure proposed in the CFD literature for discretisation error assessment [6, 27]. The drag force was selected as the reference variable for the convergence analysis, as it represents the primary quantity of interest in the present aerodynamic investigation.

Three systematically refined meshes (coarse, medium, and fine) were employed. Let f_1 ,

f_2 , and f_3 denote the drag force values obtained on the fine, medium, and coarse meshes, respectively, and let r_{21} and r_{32} be the grid refinement ratios defined as:

$$r_{21} = \left(\frac{N_1}{N_2} \right)^{1/3}, \quad r_{32} = \left(\frac{N_2}{N_3} \right)^{1/3},$$

where N_1 , N_2 , and N_3 are the total number of cells of the fine, medium, and coarse meshes.

The apparent order of convergence p was estimated by solving the nonlinear equation:

$$p = \frac{\ln \left| \frac{f_3 - f_2}{f_2 - f_1} \right|}{\ln(r_{32})},$$

yielding:

$$p = 1.18.$$

Although lower than the formal second-order accuracy of the spatial discretisation scheme, this value is consistent with external aerodynamic simulations involving complex geometries, unstructured meshes, and turbulence modelling, where local mesh non-uniformity and non-linear flow features typically reduce the observed convergence order [4, 6].

Based on the estimated order of convergence, Richardson extrapolation was used to approximate the asymptotic value of the drag force:

$$f_{\text{ext}} = f_1 + \frac{f_1 - f_2}{r_{21}^p - 1},$$

resulting in:

$$F_{\text{ext}} = 36.70 \text{ N}.$$

The Grid Convergence Index was then computed for the fine–medium and medium–coarse mesh pairs as:

$$\text{GCI}_{21} = F_s \frac{|f_1 - f_2|}{f_1(r_{21}^p - 1)}, \quad \text{GCI}_{32} = F_s \frac{|f_2 - f_3|}{f_2(r_{32}^p - 1)},$$

where $F_s = 1.25$ is a safety factor recommended for comparisons involving three grids.

The resulting GCI values were:

$$\text{GCI}_{21} = 0.63\% \quad (\text{fine–medium}), \quad \text{GCI}_{32} = 1.94\% \quad (\text{medium–coarse}).$$

The monotonic decrease of the GCI values with increasing mesh resolution, together with an asymptotic range indicator close to unity, confirms that the simulations lie within the asymptotic range of convergence. This demonstrates that discretisation errors are sufficiently small and well controlled.

Given the low value of GCI_{21} and the substantial increase in computational cost associated with the fine mesh, the medium mesh (approximately 3.2×10^7 cells) was selected for all subsequent simulations presented in this work. This choice provides an appropriate compromise between numerical accuracy and computational efficiency, ensuring that the aerodynamic differences discussed in the results chapter are not dominated by mesh-induced numerical uncertainty.

3.3.7. Computational Cost and High-Performance Computing Infrastructure

All CFD simulations were executed on the CFD HUB high-performance computing cluster of Politecnico di Milano. The cluster provides dedicated computational resources optimised for large-scale CFD workloads, enabling the execution of simulations involving tens of millions of cells within reasonable turnaround times.

Each simulation was run in parallel using 16 CPU cores. The SIMPLE algorithm was employed in steady-state mode, and all cases were advanced for a total of 2500 iterations.

Based on convergence monitoring of aerodynamic forces and residuals, statistical averaging was performed over the last 500 iterations, once a stable oscillatory regime had been reached. This approach ensured that the reported aerodynamic quantities were not influenced by transient solver behaviour or initialisation effects.

The computational cost associated with each simulation was approximately 12 hours.

The medium mesh, selected following the grid convergence analysis, represented a balanced compromise between numerical accuracy and computational efficiency, allowing the full set of configurations to be simulated within the available resources.

The use of a consistent hardware platform and identical numerical settings across all simulations further ensured that differences in computational performance and convergence behaviour did not introduce bias into the comparative aerodynamic analysis.

3.4. Summary of CFD Methodology

This chapter has presented the complete computational framework adopted for the aerodynamic analysis of the Paralympic tandem configurations investigated in this study. The numerical methodology was developed with particular attention to robustness, repeatability, and sensitivity to geometric variations, given the strong dependence of tandem aerodynamics on rider posture and inter-athlete interaction [19].

A full-domain CFD approach was employed to capture asymmetric flow features and wake interactions, supported by a carefully designed computational domain and a structured refinement strategy. The mesh resolution was selected based on a dedicated grid convergence analysis conducted on the reference configuration (v0), ensuring that numerical uncertainty remained small compared to the aerodynamic differences observed between the tested configurations.

Boundary conditions, turbulence modelling, solver settings, and convergence criteria were defined consistently across all simulations, while computational cost considerations guided the selection of a medium-resolution mesh compatible with large-scale parametric analysis on a high-performance computing platform.

The methodological choices described in this chapter provide a solid foundation for the aerodynamic results presented in chapter 5, where the influence of rider posture and relative positioning on aerodynamic performance will be analysed and discussed in detail.

4 | Biomechanical Analysis

Methodology

4.1. Overview of Biomechanical Experimental Design

The aerodynamic optimisation of tandem time-trial positions cannot be evaluated independently from their biomechanical feasibility. While aerodynamic gains are essential in time-trial performance, they often require aggressive postures that may increase muscular demand, reduce comfort, or accelerate the onset of fatigue.[11, 32]

For this reason, the present study includes an experimental biomechanical investigation aimed at assessing the muscular implications of different stoker postures corresponding to the aerodynamic configurations analysed in the CFD part of the thesis. The objective of this analysis is not the absolute optimisation of cycling biomechanics, but rather a comparative evaluation of different riding positions, with particular emphasis on muscle activation patterns and fatigue-related effects.

All biomechanical measurements were conducted on a single elite Paralympic stoker (male, 26 years old; height 1.70 m; body mass 65 kg) under controlled laboratory conditions at the Politecnico di Milano (Lecco campus). The athlete presents a visual impairment and competes in the UCI Para-cycling Category B, which includes visually impaired riders competing on tandem bicycles with a sighted pilot. This classification is specific to visual impairment and does not involve any other physical limitations, a distinction that is relevant for the interpretation of biomechanical and EMG data.

All experimental procedures were carried out in accordance with the ethical standards of the Politecnico di Milano and complied with institutional regulations for research involving human participants. Written informed consent was obtained prior to data collection.

The decision to focus exclusively on the stoker reflects both practical and scientific considerations. From a practical standpoint, the stoker's position is less constrained by UCI regulations and does not require bicycle control or manoeuvring, allowing greater freedom

in posture adjustment compared to the pilot. From a scientific perspective, modifications in stoker posture are expected to exert a significant influence on aerodynamic interaction effects within the tandem system.

Three distinct riding positions were investigated: a road-like posture, a frame-clench posture, and a time-trial posture using aerobars. These positions, visible in Figure 4.1, correspond to specific aerodynamic configurations investigated in the CFD analysis and represent realistic setups used in competition or training. For each position, cycling trials were performed at controlled power output and cadence levels, including prolonged efforts designed to induce muscular fatigue.

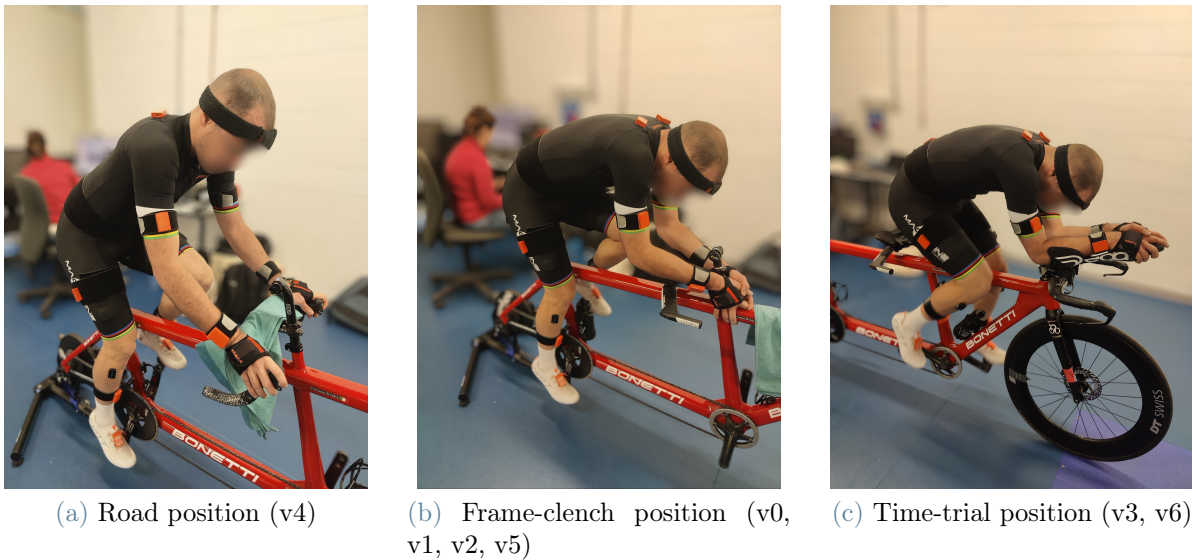


Figure 4.1: Riding positions analysed in the biomechanical experiments and their correspondence with the aerodynamic configurations used in the CFD analysis.

During each trial, full-body kinematics and muscle activity were recorded simultaneously. Kinematic data were acquired using a full-body inertial motion capture system, while muscular activity was measured through surface electromyography. This combined approach allows the integration of kinematic information with direct experimental measurements of muscle activation, providing a robust basis for biomechanical comparison between postures.

The acquired motion data were subsequently imported into *AnyBody Technology*, a musculoskeletal modelling environment, to perform kinematic and inverse dynamic analyses, as schematized in Figure 4.2.

This combined dataset enables a direct comparison between experimentally measured neuromuscular activity and musculoskeletal model outputs, which is addressed in Chapter 6.

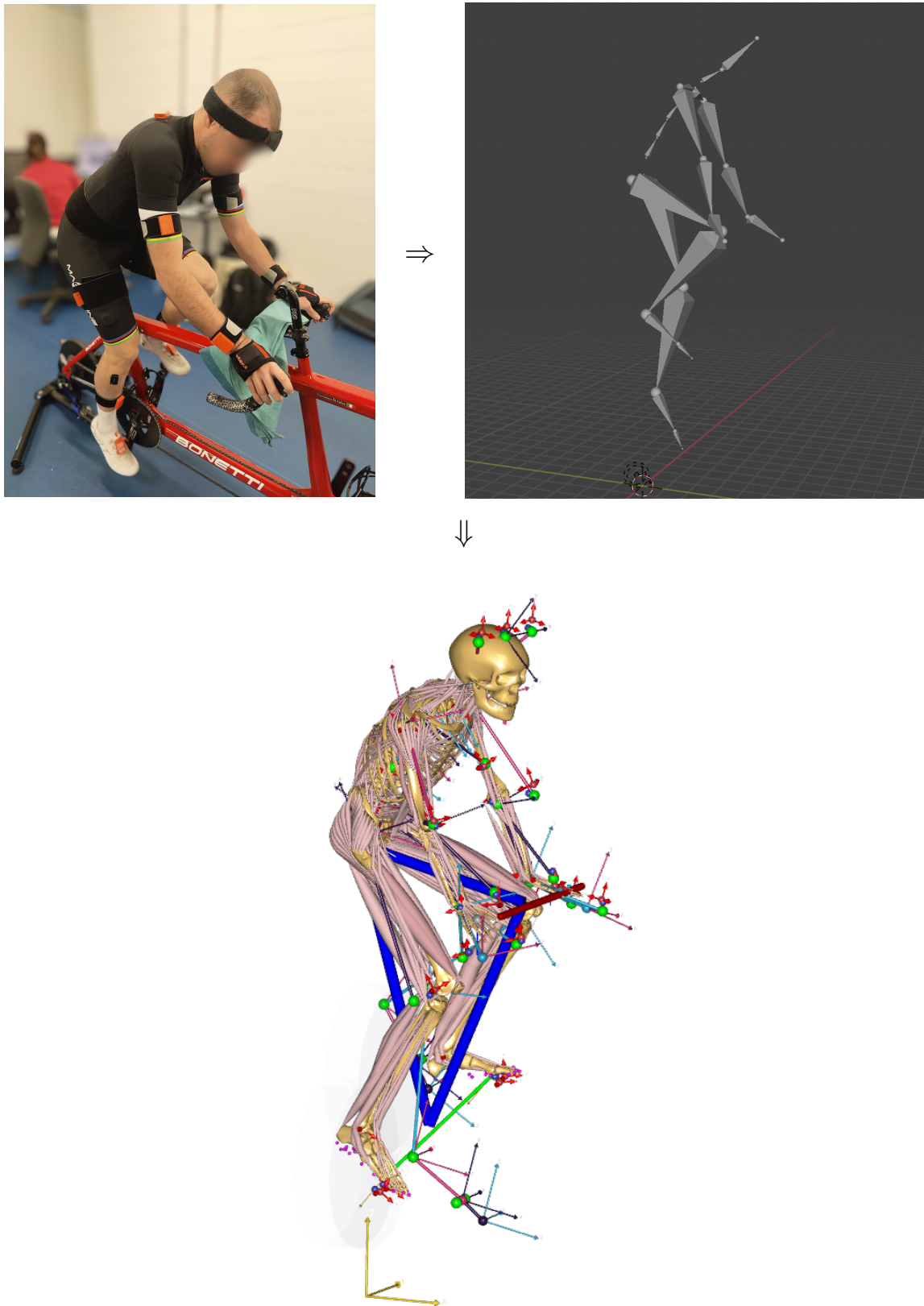


Figure 4.2: Biomechanical workflow adopted in this thesis: experimental motion and EMG acquisition, kinematic processing, and musculoskeletal simulation in the AnyBody Modeling System.

Overall, this experimental design enables a systematic assessment of the biomechanical implications of aerodynamic-driven posture modifications, supporting the integrated aero–biomechanical perspective adopted throughout the thesis.

4.2. Experimental Setup and Protocol

4.2.1. Test Environment and Equipment

All biomechanical experiments were conducted in a controlled laboratory environment at the Politecnico di Milano, Lecco campus. The tandem bicycle was mounted on a *Elite Direto* indoor training device, allowing free pedalling while maintaining a stable and repeatable experimental configuration. This setup enables realistic cycling mechanics while ensuring safety and repeatability across trials.

The tandem bicycle used in the experiments was the same platform adopted during competition. Depending on the riding position under investigation, different cockpit configurations were mounted for the stoker, including a standard road handlebar and a time-trial handlebar with aerobars.

Power output and cadence were measured using *Favero Assioma Duo* dual-sided power meter pedals and recorded via a *Garmin Edge 830* head unit. Real-time performance data were continuously monitored, and instantaneous values of power and cadence were verbally communicated to the visually impaired rider to provide feedback and ensure adherence to the prescribed experimental protocol.

Full-body kinematics were acquired using a *Xsens* inertial motion capture system configured for full-body acquisition. Sensors were securely attached to the athlete according to the standard Xsens MVN protocol, ensuring consistent sensor placement across all trials. The use of inertial sensors allows accurate reconstruction of cycling kinematics while preserving unconstrained pedalling motion [7]. The sensors position follows Xsens standard protocol and is represented in Figure 4.3.



Figure 4.3: Xsens sensors placement on the body. Credit: Xsens.

Muscle activity was recorded simultaneously using a *Delsys Trigno* wireless EMG system. Surface EMG sensors were placed on the stoker's lower-limb muscles selected for analysis, following established practices in cycling EMG studies [12]:

- Soleus,
- Tibialis anterior,
- Rectus femoris,
- Semitendinosus.

The use of simultaneous kinematic and EMG acquisition enables a direct temporal alignment between joint kinematics and muscle activation patterns, which is essential for cycle-based analysis in cycling biomechanics.

4.2.2. Calibration Procedure

A dedicated calibration procedure for the inertial motion capture system was performed before each change of riding position. This step is required by the Xsens system and is crucial to ensure accurate reconstruction of joint kinematics and to limit the accumulation of drift over time.

The standard Xsens calibration protocol consisted of:

- a static A-pose held for 5 seconds, with the athlete standing upright, arms slightly

abducted and relaxed along the sides, used to initialise sensor-to-segment alignment;

- 10 seconds of free walking, allowing the system to refine joint axis estimation and segment orientation.

This calibration sequence was repeated before each experimental condition to ensure consistency between trials performed in different riding positions. Furthermore, all trials were performed under constant environmental conditions, with particular attention to minimising external disturbances that could affect pedalling mechanics or sensor acquisition.

4.2.3. Riding Positions Investigated

The biomechanical analysis focused exclusively on the stoker and considered three distinct riding positions representative of realistic tandem cycling configurations, visible in Figure 4.1:

- **Road position:** a conventional road-cycling posture with hands on a standard handlebar, corresponding to aerodynamic configuration v4.
- **Frame-clench position:** a compact posture in which the stoker grips the tandem frame, corresponding to aerodynamic configurations v0, v1, v2 and v5.
- **Time-trial position:** a low, forward-leaning posture using aerobars, corresponding to aerodynamic configurations v3 and v6.

These positions were selected to span a wide range of torso inclinations and upper-body support strategies, allowing the investigation of posture-dependent changes in muscle activation and fatigue development.

4.2.4. Test Protocol and Load Conditions

For each riding position, a structured sequence of cycling trials was performed. The protocol consisted of four consecutive efforts designed to explore different combinations of power output and cadence, as well as to induce muscular fatigue:

- 30 seconds at 150 W and 80 rpm,
- 30 seconds at 150 W and 100 rpm,
- 30 seconds at 300 W and 80 rpm,
- 5 minutes at 300 W and 100 rpm.

The final 5-minute trial was specifically intended to bring the athlete into a state of

sustained muscular effort, enabling the investigation of fatigue-related effects under conditions representative of high-intensity time-trial cycling. Adequate rest was provided between trials and between riding positions to minimise carry-over fatigue.

4.2.5. Data Selection for Analysis

The selected trials were segmented into individual pedalling cycles based on the crank angle computed from the musculoskeletal model. Cycle-averaged quantities and their variability were then evaluated as functions of the pedalling phase, providing a robust basis for comparison between riding positions [16].

For completeness, biomechanical measurements were acquired bilaterally on both the right and left legs of the stoker. Both experimental EMG data and musculoskeletal model outputs were therefore available for each limb, enabling bilateral analysis while preserving a unified representation of the pedalling cycle.

4.3. Musculoskeletal Modelling in AnyBody

4.3.1. Overview of the AnyBody Modeling System

The AnyBody Modeling System (AMS) is a comprehensive software environment for the simulation of human movement, capable of computing joint reaction forces, muscle activations, mechanical work, and internal load distributions based on prescribed kinematics and external constraints. At its core, AMS employs a detailed multi-body representation of the human musculoskeletal system, combining rigid-body dynamics with anatomically accurate musculoskeletal models [9].

Each model is constructed from a set of rigid segments connected by kinematic joints (spherical, revolute, universal, or user-defined), and actuated by Hill-type muscle models. These muscles are arranged into functional groups corresponding to anatomical structures (hamstrings, quadriceps, hip flexors, gastrocnemius, etc.), each with physiologically meaningful parameters such as optimal fiber length, maximal isometric force, moment arms, and pennation angles [38].

AMS supports both forward and inverse dynamic analyses. In this study, inverse dynamics is used: the recorded motion is imposed on the digital model, and the system computes the internal forces required to reproduce it. This methodology is particularly suitable for cycling, where the kinematic pattern is highly repeatable and externally driven by the crank trajectory, allowing robust extraction of joint torques and muscle recruitment [12].

4.3.2. Importing Xsens BVH Kinematics into AnyBody

The Xsens motion capture session produced a set of time-dependent joint rotations and segment orientations exported in `.bvh` format. AnyBody provides built-in import functionality for BVH files, mapping each recorded joint rotation to the corresponding segment in the musculoskeletal model.

The import process proceeds through the following steps:

1. **Hierarchy mapping:** The BVH file defines a hierarchical structure of joints and bones. This structure is mapped onto the AnyBody model, which uses a similar tree-based representation.
2. **Rotation application:** AnyScript drivers are generated to impose the recorded joint angles frame-by-frame on the human model. These drivers act as prescribed kinematic constraints during the inverse dynamic computation.
3. **Time normalisation:** The BVH motion is converted to the AnyBody time domain, typically sampling at 100–240 Hz depending on the original acquisition.
4. **Scaling of anthropometry:** AnyBody uses a parametric human model that must be scaled to the athlete's anthropometric measurements (segment lengths, mass distribution, joint widths). A scaling law adjusts the musculoskeletal model to match the athlete's height and limb proportions.

This process ensures that the motion recorded from the athlete is reproduced with high fidelity inside AMS. The resulting kinematic chain respects not only the externally observed joint angles but also the anatomical constraints predefined in the underlying musculoskeletal model. Figure 4.4 represents the output of the motion capture process.

4.3.3. Structure of the Human Model: Joints, Segments, and Muscle Architecture

The musculoskeletal model used in AnyBody consists of [1]:

- 40–50 rigid body segments (depending on the level of detail chosen for the model),
- 100+ kinematic degrees of freedom,
- 900–1200 individual muscle-tendon units,
- multi-muscle groups sharing moment arms across the main joints.

The joint types include:

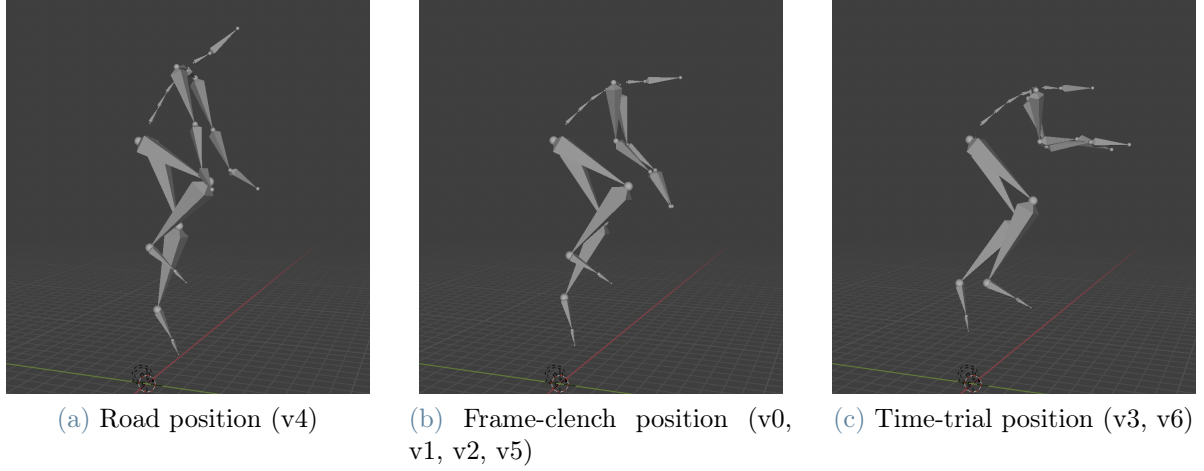


Figure 4.4: Results of motion capture in different positions, visualized in Blender

- *Spherical joints* (e.g., shoulder, hip) allowing three rotational degrees of freedom,
- *Revolute joints* (e.g., knee, elbow) for hinge-like motion,
- *Universal joints* (e.g., ankle complex),
- and *custom constraint joints* representing complex articulations such as the lumbar spine.

Each muscle is represented as a force-generating actuator characterised by a set of physiological and mechanical parameters. These include:

$$F_{\max}, \quad l_{\text{opt}}, \quad v_{\max}, \quad \alpha, \quad c(\theta),$$

where:

- $F_{\max}$ is the maximum isometric force that the muscle can generate under optimal conditions, and primarily reflects muscle physiological cross-sectional area;
- l_{opt} is the optimal muscle fibre length at which force generation is maximal, defining the peak of the force-length relationship;
- $v_{\max}$ is the maximum shortening velocity of the muscle fibres, governing the force-velocity relationship during concentric contractions;
- α is the pennation angle, defined as the angle between muscle fibres and the line of action of the tendon, which affects the transmission of fibre force to the tendon;
- $c(\theta)$ represents the muscle activation dynamics, describing the relationship between

neural excitation and muscle activation as a function of joint configuration and time.

Moment arms are defined as functions of joint angles, allowing the dynamic computation of torque capacity throughout the motion cycle. This is crucial for cycling, where hip and knee angles vary significantly across the pedal stroke, continuously modulating the muscle force-length relationship. [1]

4.3.4. Mathematical Formulation of Inverse Dynamics

In the AnyBody Modeling System, inverse dynamics is formulated as the solution of the equations of motion of a constrained multibody system for each time frame of the recorded movement. Given the measured kinematics, the model enforces dynamic equilibrium of all body segments while satisfying the kinematic constraints imposed by the bicycle–rider interaction.

For a generic configuration, mechanical equilibrium can be expressed in compact form as

$$\mathbf{C}\mathbf{f} + \mathbf{R} = \mathbf{Q},$$

where $\mathbf{f}$ is the vector of muscle forces, $\mathbf{C}$ is the corresponding moment arm matrix, $\mathbf{R}$ represents joint reaction and constraint forces, and $\mathbf{Q}$ collects the generalized external and inertial loads required to reproduce the measured motion. In this formulation, joint torques are not prescribed quantities but arise implicitly from the balance of forces and moments acting on the system.

Because the musculoskeletal system is highly redundant, with a large number of muscle actuators acting on a comparatively limited set of degrees of freedom, the inverse dynamics problem does not admit a unique muscle force solution. Infinitely many muscle force distributions can satisfy the equilibrium equations at a given time frame.

To resolve this redundancy, muscle recruitment in AnyBody is formulated as an optimisation problem. In the present study, muscle forces were computed using the default recruitment strategy implemented in the AnyBody Modeling System, which is based on a min/max criterion [26]. This approach minimises the maximum relative muscle activation across all muscles, leading to a physiologically plausible load sharing without the introduction of user-defined weighting parameters.

The optimisation is solved independently at each time step, resulting in time-varying muscle forces, muscle activations, and joint reaction forces. These outputs provide a consistent mechanical description of the movement but should be interpreted as the outcome

of a modelling-based optimality assumption rather than as a direct representation of the subject's true neuromuscular control strategy.

While the min/max recruitment criterion has been shown to provide physiologically reasonable muscle force distributions in simulations of daily activities and standard locomotor tasks in healthy adults, its validity under high-intensity athletic conditions is less certain. In particular, during maximal or near-maximal efforts, trained athletes may adopt neuromuscular strategies that deviate from the assumption of minimising the maximum relative activation. Elite performers often exhibit task-specific motor adaptations, selective co-contractions, and intentional redistribution of load that may not be fully captured by a generic optimisation criterion.

Therefore, the predicted muscle forces and activations should be interpreted as mechanically consistent solutions within the framework of the adopted optimality principle, rather than as a definitive representation of the athlete's true neural control strategy. This limitation is especially relevant in high-performance cycling, where force production approaches physiological limits and motor coordination is shaped by long-term training adaptations.

4.3.5. Constraint-Based Modelling of the Bicycle

Cycling is characterised by strong kinematic constraints that couple body motion with the bicycle. In AnyBody, these constraints were implemented as follows:

Foot–pedal constraint. Each foot is attached to its pedal using a kinematic spherical constraint. This ensures that the ankle, knee, and hip follow the required crank-driven motion while allowing natural rotation of the foot around the pedal spindle.

Pelvis–saddle constraint. The interaction between the pelvis and the saddle was modelled using a mixed constraint formulation. The pelvis was connected to the saddle through a soft kinematic constraint that enforces contact while allowing limited relative motion.

Specifically, a reaction force was applied in the direction normal to the saddle surface, preventing penetration and ensuring continuous contact. In the tangential directions, frictional constraints were introduced to resist relative sliding between the pelvis and the saddle, while still allowing small translations consistent with real-world saddle compliance.

Hand–handlebar constraint. Hands were constrained to the handlebars through spherical joints with limited motion, reproducing the stabilising function of upper limbs in

supporting and positioning the torso.

These constraints force the human model to execute the pedalling motion realistically while ensuring that the resulting joint torques arise from physiologically valid movement patterns. This model was adapted from a previously existing model for the simulation of bicycle pedalling motion, developed by AnyBody Technology for Vicon motion capture systems, available in the AnyBody Models Repository [1].

4.3.6. Bicycle Geometry Optimisation

Before running the inverse dynamics, AnyBody performs an optimisation step to adjust bicycle model parameters so that the imported BVH motion matches all applied kinematic constraints. Variables include:

- saddle height,
- saddle setback,
- cranks position,
- handlebar reach and drop,

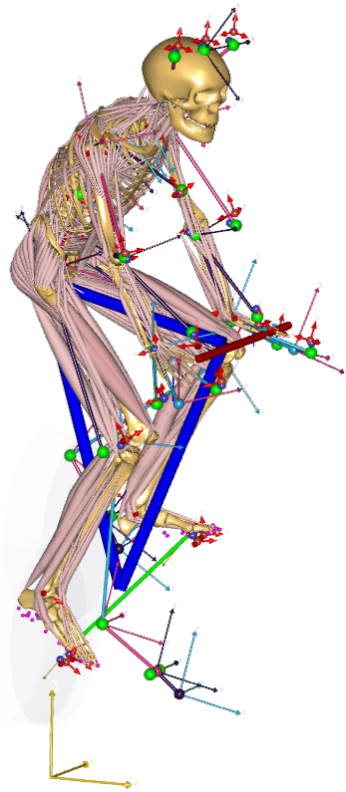
ensuring precise matching between the virtual bicycle and the real tandem setup used during acquisition. This optimisation ensures that constraints do not introduce artificial forces.

4.3.7. Running Dynamic and Muscle Recruitment Analysis

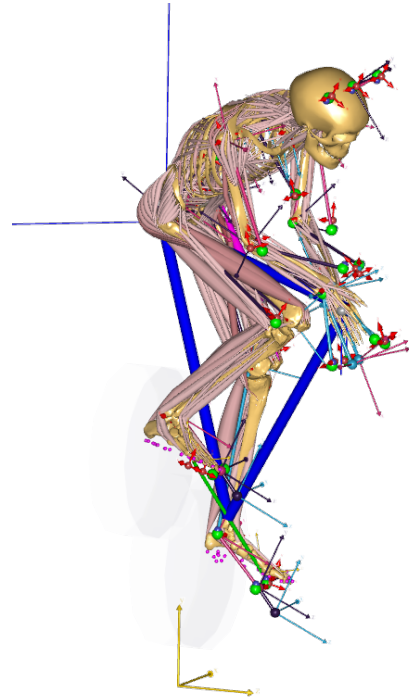
The analysis is performed over several pedal cycles. At each time frame:

1. joint kinematics are imposed,
2. reaction forces at the pedals, saddle, and handlebar are computed,
3. inverse dynamics yields net joint torques,
4. optimisation allocates torques across muscles,
5. muscle activations and forces are recorded.

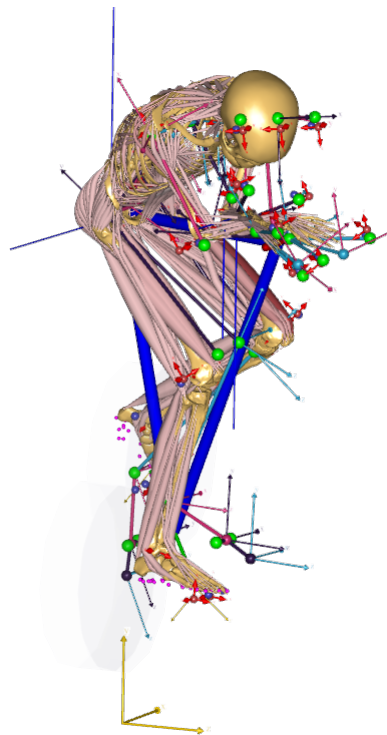
The inverse dynamics analysis provides a set of time-dependent internal variables describing the mechanical and neuromuscular demands associated with the prescribed motion.



(a) Road position (v4)



(b) Frame-clench position (v0, v1, v2, v5)



(c) Time-trial position (v3, v6)

Figure 4.5: Results of the musculoskeletal simulations for the different stoker riding positions.

The main output quantities include:

$$F_{\text{muscle}}(t), \quad a(t), \quad \boldsymbol{\tau}_{\text{joint}}(t), \quad \mathbf{R}_{\text{joint}}(t),$$

where:

- $F_{\text{muscle}}(t)$ denotes the muscle force generated by each muscle–tendon unit as a function of time;
- $a(t)$ represents the muscle activation level, a dimensionless quantity typically bounded between 0 and 1, describing the neural excitation required to produce the computed muscle force;
- $\boldsymbol{\tau}_{\text{joint}}(t)$ are the net joint torques resulting from the combined action of muscles, external loads, and kinematic constraints;
- $\mathbf{R}_{\text{joint}}(t)$ are the joint reaction forces, representing the internal forces transmitted across the joint interfaces to maintain mechanical equilibrium.

4.3.8. Temporal Scope of the AnyBody Simulations

Due to the high computational cost associated with full musculoskeletal inverse-dynamics simulations, the AnyBody analyses were not performed over the entire duration of all experimental trials.

Musculoskeletal simulations were carried out exclusively for the 5-minute cycling trial performed at 300 W and 100 rpm, which represents the most demanding condition in terms of mechanical load and is the primary reference condition for high-intensity time-trial cycling.

Furthermore, within this trial, the inverse dynamic analysis was limited to a selected temporal window corresponding to a few seconds of steady-state pedalling, including approximately 30 pedalling cycles. This window was chosen to be representative of regular, cycle-to-cycle pedalling behaviour, excluding transient phases such as acceleration or cadence adjustments.

As a result, AnyBody outputs are available only for a limited number of consecutive pedalling cycles. The implications of this restricted temporal scope for the comparison with experimentally measured EMG data are discussed in Chapter 6.

4.4. Kinematic Analysis

4.4.1. Scope and Rationale

In addition to muscle activation and fatigue analysis, a kinematic analysis was performed to quantify posture-related differences between riding positions and to assess the capability of the musculoskeletal model to reproduce cycling kinematics.

Kinematic analysis was conducted exclusively on the 5-minute cycling trial at 300 W and 100 rpm for each riding position. This condition represents steady-state pedalling under high mechanical load and provides a consistent basis for posture comparison. Unlike muscle activation or fatigue metrics, joint kinematics are expected to remain stable over time during steady cycling, making the analysis of a single representative trial sufficient.

4.4.2. Postural Angles

To characterise the overall riding posture, mean values and standard deviations of selected joint angles were computed over the entire analysed time window. The following quantities were considered:

- trunk inclination angle with respect to the horizontal plane,
- shoulder joint orientation,
- elbow flexion angle.

Mean values represent the average posture adopted during pedalling, while the corresponding standard deviations provide an indication of postural stability throughout the cycle. These metrics allow a direct comparison of torso and upper-body configuration across the three investigated riding positions [24, 36].

4.4.3. Range of Motion Analysis

To quantify cyclic joint motion during pedalling, the range of motion (ROM) was computed for the hip, knee, and ankle joints of both legs. For each joint, ROM was defined as the difference between the maximum and minimum joint angle observed during the analysed interval:

$$\text{ROM} = \theta_{\max} - \theta_{\min}.$$

ROM values provide insight into how different riding positions influence joint excursion and potential compensatory strategies adopted to accommodate postural constraints [3]. Bilateral ROM analysis further allows the identification of asymmetries between the right and left legs.

The results of kinematic analysis are discussed in Chapter 6.

4.5. Fatigue Analysis Based on EMG Measurements

4.5.1. Rationale and Overview

Muscular fatigue is a key determinant of cycling performance, especially under sustained high-intensity conditions typical of time-trial efforts. During prolonged cycling, fatigue-related adaptations can manifest both as changes in the overall activation amplitude required to maintain the target workload and as changes in the frequency content of the EMG signal, which is often associated with peripheral fatigue mechanisms (e.g., reduced muscle fibre conduction velocity) [10].

Given the cyclic and dynamic nature of pedalling, fatigue assessment in this study was performed using complementary EMG-based approaches:

- **Time-based analysis** using sliding windows across the entire 5-minute trial to track gradual fatigue-related trends;
- **Cycle-based analysis** computing fatigue indicators cycle-by-cycle.

This dual strategy follows common recommendations in cycling EMG analysis, where both amplitude-related metrics (e.g., RMS) and spectral metrics (e.g., median/mean frequency) are used to characterise fatigue [12, 16].

4.5.2. Data Selection

Fatigue analysis was conducted on the 5-minute cycling trials performed at 300 W and 100 rpm for each riding position (road, frame-clench, time-trial). This condition provides sufficient duration to observe fatigue-related adaptations while maintaining a workload representative of high-intensity time-trial cycling.

EMG recordings were available bilaterally for the selected muscles: tibialis anterior (TA), semitendinosus (ST), rectus femoris (RF), and soleus (SOL). Bilateral analysis allows the identification of potential asymmetries and supports internal consistency checks between right and left limbs.

4.5.3. EMG Pre-processing

Raw EMG signals were pre-processed using a fourth-order zero-phase Butterworth band-pass filter in the range 20–450 Hz to reduce motion artefacts and high-frequency noise while preserving the physiological EMG bandwidth [10]. Filtered signals were de-measured prior to subsequent computations.

4.5.4. Fatigue Metrics

Fatigue indicators were computed using both *time-domain* (amplitude) and *frequency-domain* descriptors.

Amplitude metric (RMS). For each analysis segment (time window or pedalling cycle), the root mean square (RMS) was computed as:

$$\text{RMS} = \sqrt{\frac{1}{N} \sum_{n=1}^N x[n]^2},$$

where $x[n]$ is the band-pass filtered EMG and N is the number of samples in the segment. RMS is widely used in pedalling EMG analysis as a measure of muscle activity level and neural drive [16].

Spectral metrics (MF and MNF). For each segment, the power spectral density (PSD) was estimated using Welch’s method. Metrics were computed within the 20–450 Hz band:

- **Median frequency (MF):** the frequency f_{50} such that

$$\sum_{f \leq f_{50}} P(f) = \frac{1}{2} \sum_f P(f),$$

where $P(f)$ is the PSD.

- **Mean frequency (MNF):**

$$\text{MNF} = \frac{\sum_f f P(f)}{\sum_f P(f)}.$$

A fatigue-related shift of spectral content toward lower frequencies is a classical EMG marker of peripheral fatigue mechanisms, although its magnitude can be task- and subject-

dependent, especially in dynamic exercise [10, 13].

4.5.5. Time-Based Analysis (Sliding Window)

To track fatigue progression across the full trial, RMS, MF, and MNF were computed in sliding windows of length $T_w = 2.0$ s with overlap $T_{ov} = 1.0$ s. For each window k , the mid-time t_k was recorded and converted to a normalised progression axis:

$$x_k = 100 \frac{t_k - t_1}{t_{end} - t_1},$$

expressed as percentage of trial duration.

This approach provides a continuous description of fatigue-related trends with a tunable balance between temporal resolution and estimator robustness, as commonly adopted in continuous fatigue monitoring frameworks [13].

4.5.6. Cycle-Based Analysis (Pedal-Cycle Segmentation)

In parallel, fatigue metrics were computed on a cycle-by-cycle basis. Pedalling cycles were detected using a composite EMG signal built from the rectified, channel-normalised EMG:

$$c(t) = \frac{1}{M} \sum_{m=1}^M \frac{|x_m(t)|}{\max(|x_m|)},$$

followed by peak detection with:

- minimum peak distance corresponding to 0.45 s,
- minimum peak prominence of 0.30,
- cadence plausibility constraint (85–115 rpm).

Each detected cycle was required to contain a minimum number of samples to exclude partial or corrupted revolutions. For every valid cycle, RMS, MF, and MNF were computed on the cycle segment.

Cycle-based analysis is well suited to cyclic tasks as it reduces sensitivity to short transients and links fatigue descriptors to the natural repetition of the motor task [12].

4.5.7. Trend Quantification and Statistical Comparison

Fatigue progression was quantified using two complementary approaches, applied to both time-based and cycle-based series:

Robust linear trend (slope). For each muscle and metric, a robust linear regression was fitted:

$$y(x) = \beta_0 + \beta_1 x,$$

where x is trial progression in % and y is the fatigue metric (RMS, MF, MNF). The slope β_1 represents the average trend over the trial while reducing sensitivity to outliers [13].

Early vs. late comparison. To provide an interpretable summary independent of linearity assumptions, the time series was split into two segments:

$$\text{early} = [0, 50\%], \quad \text{late} = (50\%, 100\%],$$

and a non-parametric rank-sum test was used to compare early vs late distributions. In addition, the percentage change from early to late was computed as:

$$\Delta y_{\%} = 100 \frac{\bar{y}_{\text{late}} - \bar{y}_{\text{early}}}{\bar{y}_{\text{early}}}.$$

These two summaries (slope and early–late difference) were used together to characterise fatigue-related changes in a way that is both robust and easy to compare across positions and muscles.

4.5.8. Bilateral Analysis and Symmetry Check

All analyses were performed bilaterally. Beyond reporting bilateral results, a simple internal consistency check was implemented by comparing right vs left trends for homologous muscles (e.g., MF slope of TA-R vs TA-L). This symmetry check does not assume perfect symmetry but helps identify gross acquisition artefacts or unexpected unilateral behaviour under controlled laboratory conditions.

4.6. Overview of Outcome Measures

Table 4.1 summarises the biomechanical analyses performed, specifying for each outcome measure the corresponding data source, riding position, and experimental trial considered.

Final results and their interpretation are presented in Chapter 6.

Table 4.1: Overview of the main biomechanical outcome measures, corresponding data sources, and experimental conditions.

Analysis	Data source	Riding positions	Trials considered
Kinematic analysis	Xsens motion capture processed in AnyBody	Road, Frame-clench, Time-trial	5 min – 300 W, 100 rpm
EMG activation patterns	Surface EMG (Delsys Trigno)	Road, Frame-clench, Time-trial	All four trials
EMG–musculoskeletal simulation comparison	Surface EMG + AnyBody inverse dynamics	Road, Frame-clench, Time-trial	5 min – 300 W, 100 rpm
Fatigue analysis	Surface EMG (frequency and amplitude-based indicators)	Road, Frame-clench, Time-trial	5 min – 300 W, 100 rpm

5 | Aerodynamic Results

5.1. Overview of Results and Comparison Metrics

This chapter presents the aerodynamic results obtained from the CFD campaign described in Chapter 3. The objective is to quantify and interpret the effects of pilot posture, stoker posture, and inter-rider spacing on the aerodynamic performance of a tandem time-trial bicycle.

The analysed configurations (v0–v8) were designed to progressively modify specific geometric features of the system, allowing both the isolation of individual effects and the assessment of interaction-driven phenomena.

The primary performance metric adopted throughout this chapter is the effective drag area:

$$C_d A = \frac{F_x}{\frac{1}{2} \rho U_\infty^2}, \quad (5.1)$$

where F_x is the streamwise aerodynamic force, ρ is the air density, and U_∞ is the inlet velocity.

Aerodynamic forces are obtained through surface integration over dedicated patches corresponding to the pilot, stoker and bicycle. Unless otherwise stated, all reported quantities correspond to time-averaged values computed over the final 500 solver iterations, once statistical convergence was achieved.

To support the physical interpretation of the results, the analysis combines:

- global and component-wise $C_d A$ comparisons,
- cumulative drag distributions along the streamwise (x) and vertical (y) directions,
- qualitative inspection of pressure and velocity fields to identify separation regions and wake interaction mechanisms.

5.2. Global Aerodynamic Performance Across Configurations

Figure 5.1 and Table 5.1 report the mean global C_dA values for configurations v0–v6 and pilot-only, together with the standard deviation computed over the averaging window.

For clarity, a visual overview of the 3D models considered in the CFD analysis is provided in Figure 5.2 to recall the investigated configurations. The v7 and v8 cases include the only pilot, without bike and stoker, to isolate the effective improvement between the two positions.

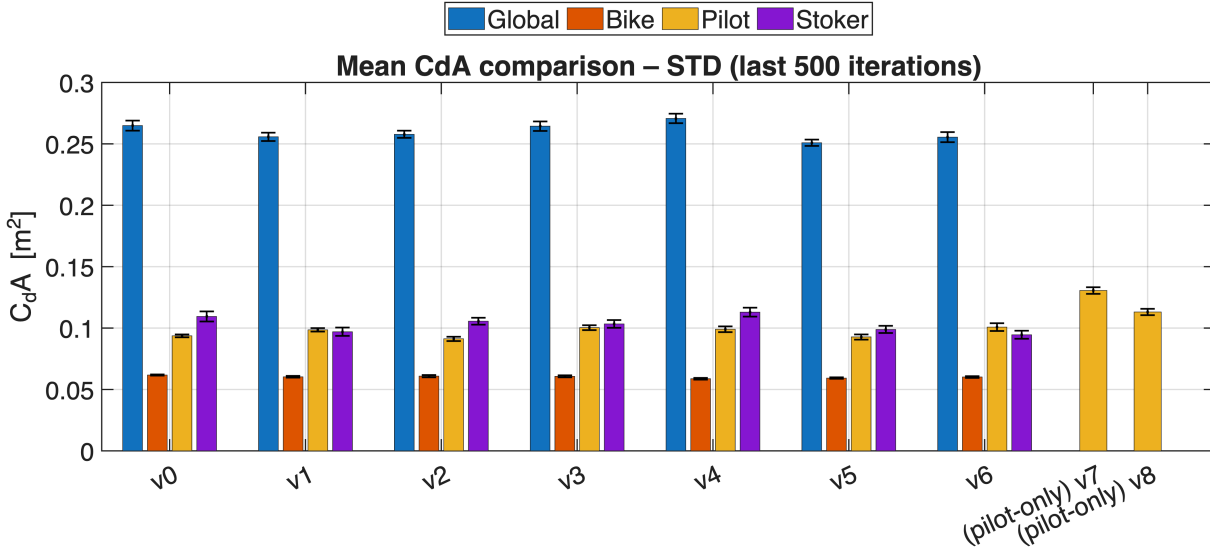


Figure 5.1: Mean global C_dA values and standard deviation over the final 500 iterations for configurations v0–v6 and pilot-only

The global drag area spans from 0.251 to 0.271 m² across tandem configurations. The bicycle contribution remains nearly constant across all configurations, indicating that the observed differences are almost entirely driven by rider aerodynamics and by aerodynamic interaction effects between pilot and stoker.

Among the tandem configurations, v5 exhibits the lowest global C_dA , whereas v3 and v4 show the highest values. The following sections provide a detailed analysis of these results.

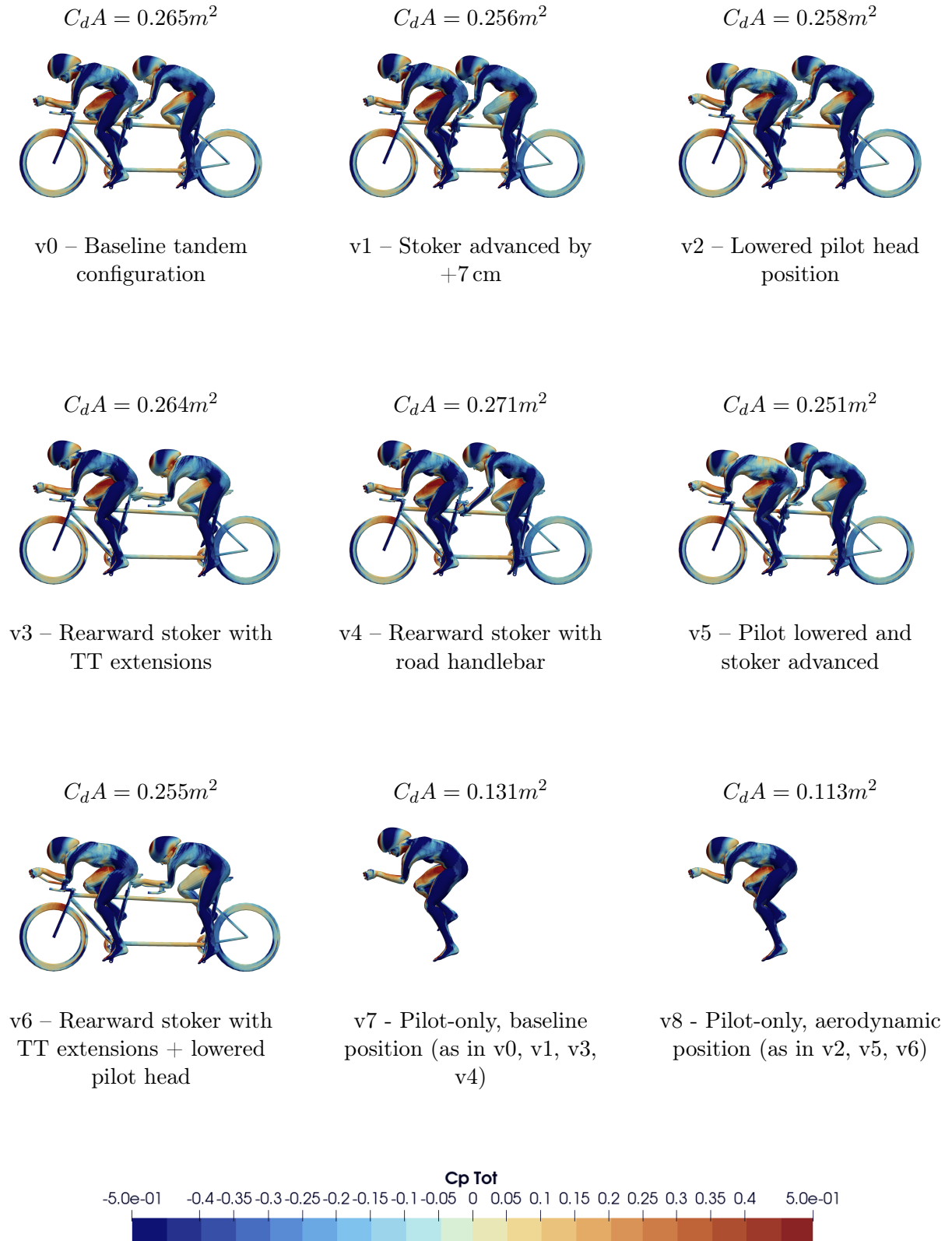


Figure 5.2: Three-dimensional CAD models of the nine configurations analysed in the CFD study (v0–v8). The mean global drag area (C_dA), computed over the last 500 solver iterations, is reported above each model for reference. The surfaces are coloured according to the total pressure coefficient ($C_{p,tot}$).

Table 5.1: Mean aerodynamic drag area C_dA (mean $\pm$ STD), computed over the last 500 iterations. The relative variation of the global drag area with respect to the baseline configuration v0 is also reported.

Config.	Global [m^2]	Bike [m^2]	Pilot [m^2]	Stoker [m^2]	$\Delta C_dA^{\text{Global}}$ [%]
v0	0.265 ± 0.004	0.062 ± 0.001	0.094 ± 0.001	0.109 ± 0.004	0.0
v1	0.256 ± 0.003	0.060 ± 0.001	0.099 ± 0.002	0.097 ± 0.003	-3.4
v2	0.258 ± 0.003	0.061 ± 0.001	0.091 ± 0.002	0.106 ± 0.003	-2.6
v3	0.264 ± 0.004	0.061 ± 0.001	0.100 ± 0.002	0.103 ± 0.003	-0.4
v4	0.271 ± 0.004	0.059 ± 0.001	0.099 ± 0.002	0.113 ± 0.004	+2.3
v5	0.251 ± 0.003	0.059 ± 0.001	0.093 ± 0.002	0.099 ± 0.003	-5.3
v6	0.255 ± 0.004	0.060 ± 0.001	0.101 ± 0.003	0.095 ± 0.003	-3.8
v7	0.131 ± 0.003	—	0.131 ± 0.003	—	-50.6
v8	0.113 ± 0.003	—	0.113 ± 0.003	—	-57.4

5.3. Effect of Pilot Head and Upper-Body Alignment

Configurations v2, v5 and v6 share the same lowered pilot head and more compact upper-body alignment. The pilot-only reference cases (v7 and v8) isolate the effect of pilot posture in the absence of tandem interaction and confirm that lowering the pilot head produces a substantial reduction in drag.

When the pilot operates within a tandem configuration, however, the effective pilot drag is not solely determined by pilot posture, but also by the aerodynamic interaction with the stoker. As a consequence, a lowered pilot head does not necessarily lead to a consistent reduction in the pilot drag contribution across all tandem configurations.

While configurations v2 and v5 benefit from both improved pilot posture and favourable interaction effects, configuration v6 exhibits an increased pilot drag despite the same head-lowered posture. This behaviour highlights that, in tandem aerodynamics, interaction-driven effects associated with inter-rider spacing and wake development can outweigh the intrinsic benefits of pilot posture optimisation.

As discussed in the following sections, wake interaction with the downstream rider can either amplify or offset the intrinsic benefit associated with pilot head lowering, depending on the resulting wake structure and pressure recovery mechanisms.

5.4. Influence of Inter-Rider Spacing and Aerodynamic Interaction

Inter-rider spacing plays a central role in determining the aerodynamic coupling between pilot and stoker. Modifying the relative longitudinal position of the riders alters both wake development and pressure recovery mechanisms, leading to non-monotonic effects on the aerodynamic loads of the two riders.

In configuration v1, where the stoker is advanced by 7 cm relative to the baseline, the pilot drag contribution increases slightly with respect to v0, despite the pilot posture remaining unchanged. This behaviour can be attributed to aerodynamic interference effects in the gap region: reducing the inter-rider distance intensifies the interaction between the pilot wake and the leading edge of the stoker, potentially accelerating the flow in the gap and increasing local suction on the rear surface of the pilot. As a result, the pilot base pressure is not necessarily increased, and a small penalty in pilot drag may occur even when the downstream rider benefits from improved shielding. A visual comparison is shown in Figure 5.3.

On the other hand, increasing the inter-rider spacing (v3 and v6) leads to a systematic increase in pilot drag. In these configurations, the pilot wake is allowed to expand and strengthen before interacting with the stoker, resulting in reduced pressure recovery on the pilot rear surface and a higher pressure-drag contribution, as visible in Figure 5.4.

The pilot-only cumulative drag distributions shown in Figure 5.5 demonstrate that the additional drag in configurations v3 and v6 originates primarily in the rear portion of the pilot body, associated with reduced wake closure by the stoker, rather than in the front upper-body region.

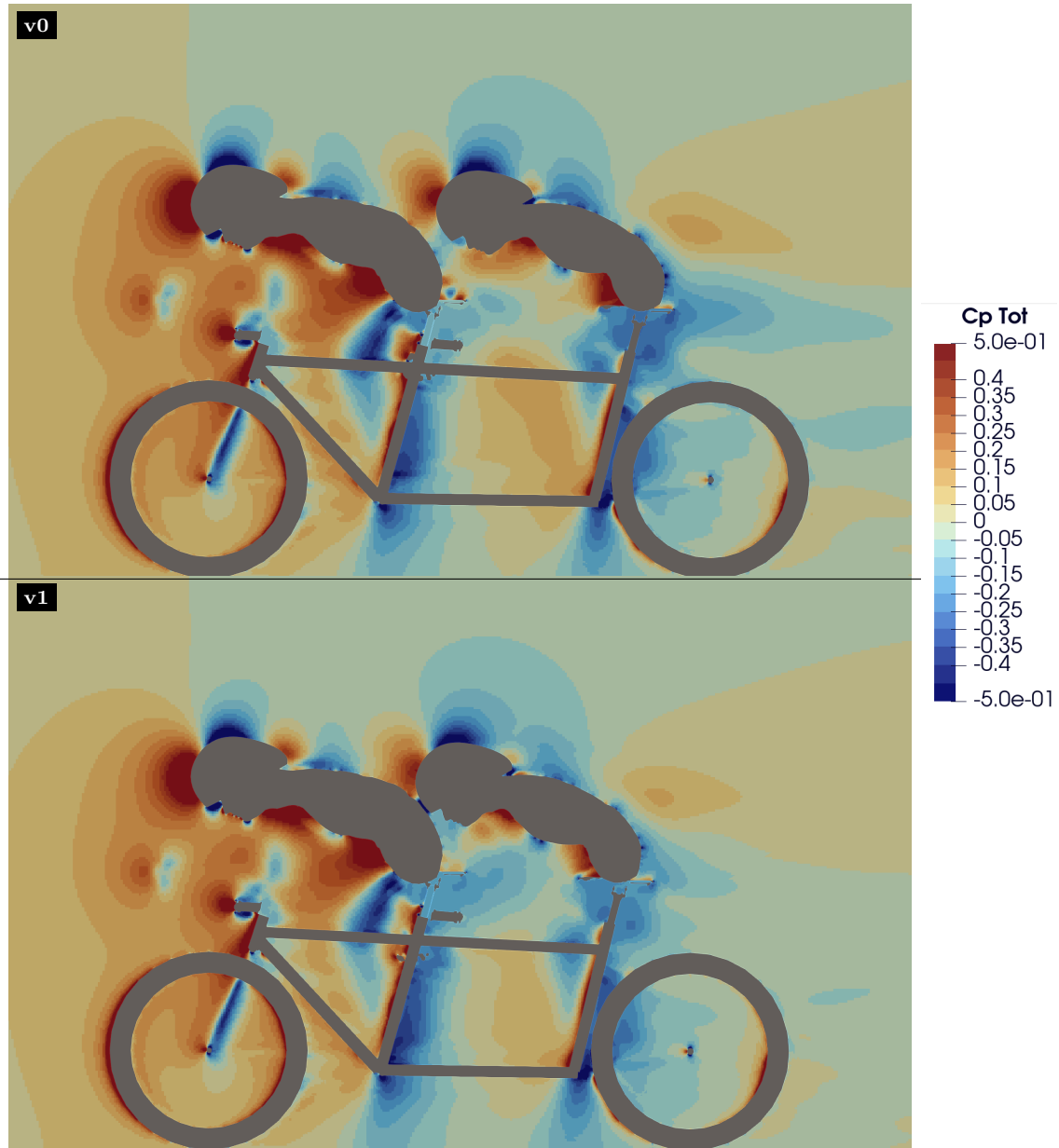


Figure 5.3: Vertical mid-plane slice of the total pressure coefficient ($C_{p,tot}$) for configurations v0 and v1. The slice is extracted along the longitudinal mid-plane of the computational domain and highlights differences in pressure distribution and wake structure between the two configurations.

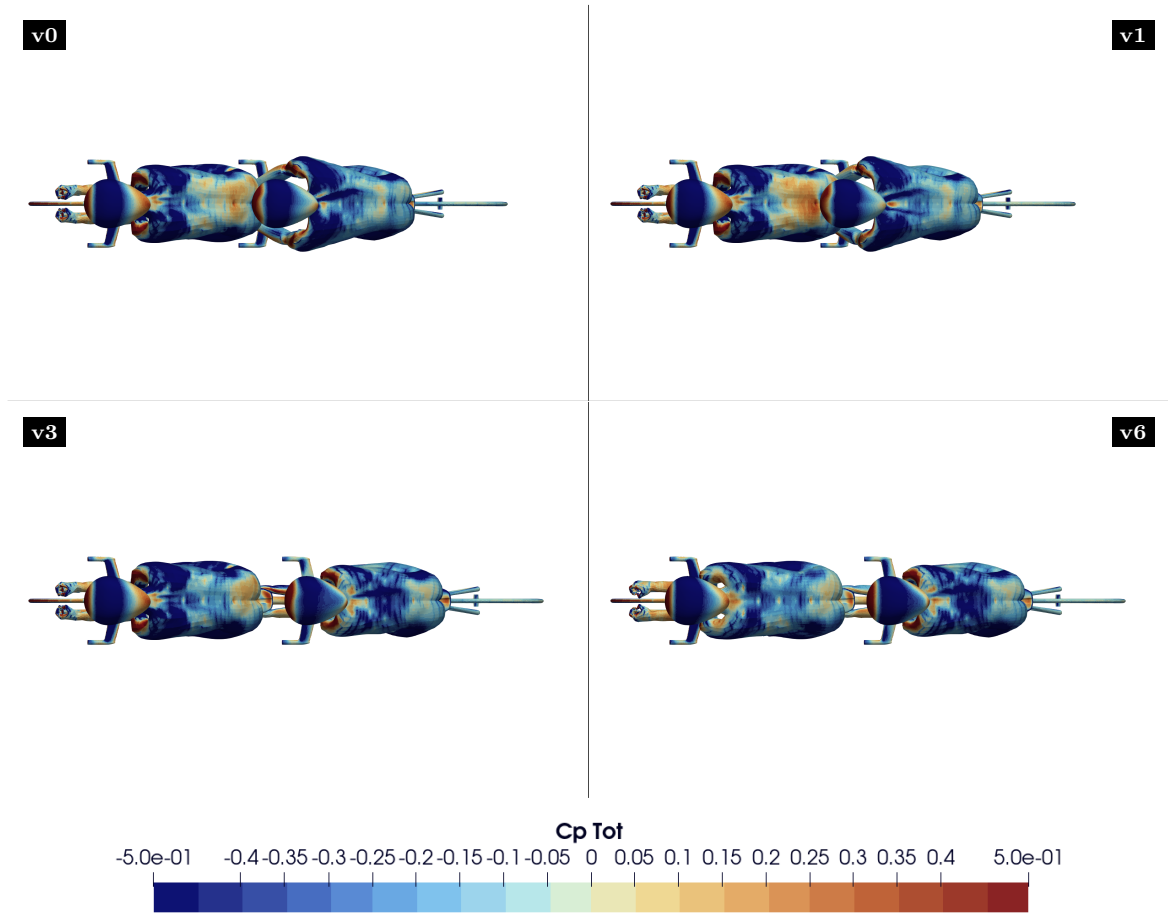


Figure 5.4: Surface total pressure coefficient ($C_{p,tot}$) distribution for the selected configurations, shown in top view. The maps highlight differences in high- and low-pressure regions over the riders. These configurations were selected as best representing the inter-rider spacing differences.

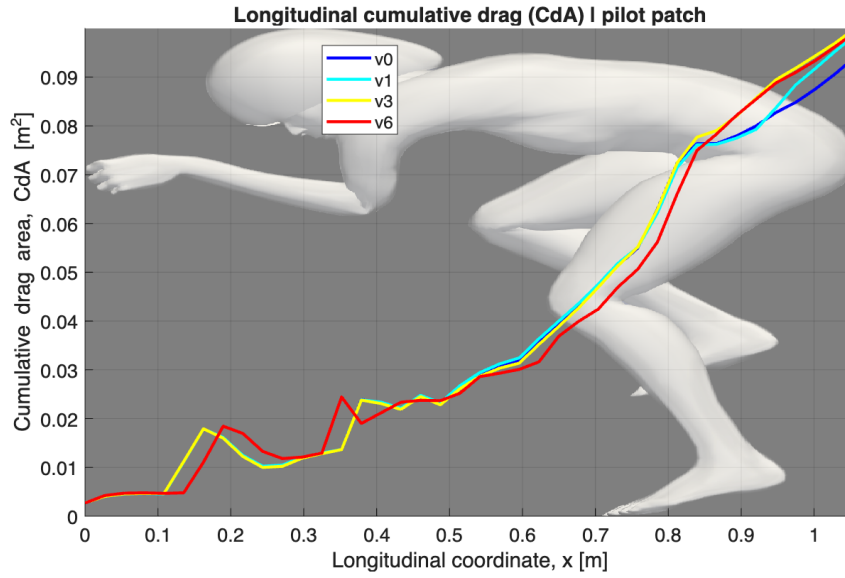


Figure 5.5: Longitudinal cumulative drag area (C_dA) on the pilot surface patch, obtained by integrating aerodynamic forces along the streamwise direction from the foremost point of the rider.

5.5. Influence of Stoker Posture and Cockpit Configuration

The stoker posture and cockpit configuration affect the local flow around the arms, shoulders and upper torso, influencing both frontal area and separation sensitivity. Time-trial extensions (v3 and v6) reduce the projected frontal area compared to a standard road handlebar (v4) and generally improve the aerodynamic efficiency of the stoker.

In configuration v3, despite the increased inter-rider spacing, the stoker drag contribution is reduced with respect to the baseline configuration v0. This result indicates that the stoker benefits from a combination of improved posture (lowered head and TT extensions) and exposure to a more uniform incoming flow, as the stoker is partially removed from the highly turbulent near-wake region of the pilot. Although the stoker experiences less shielding from the pilot, the improved flow quality and reduced separation sensitivity lead to a net reduction in stoker drag. Figure 5.6 represents the streamline visualization for these two configurations.

These results confirm that shielding alone does not determine stoker aerodynamic performance, and that posture optimisation can outweigh increased exposure to the freestream when the flow quality is sufficiently improved.

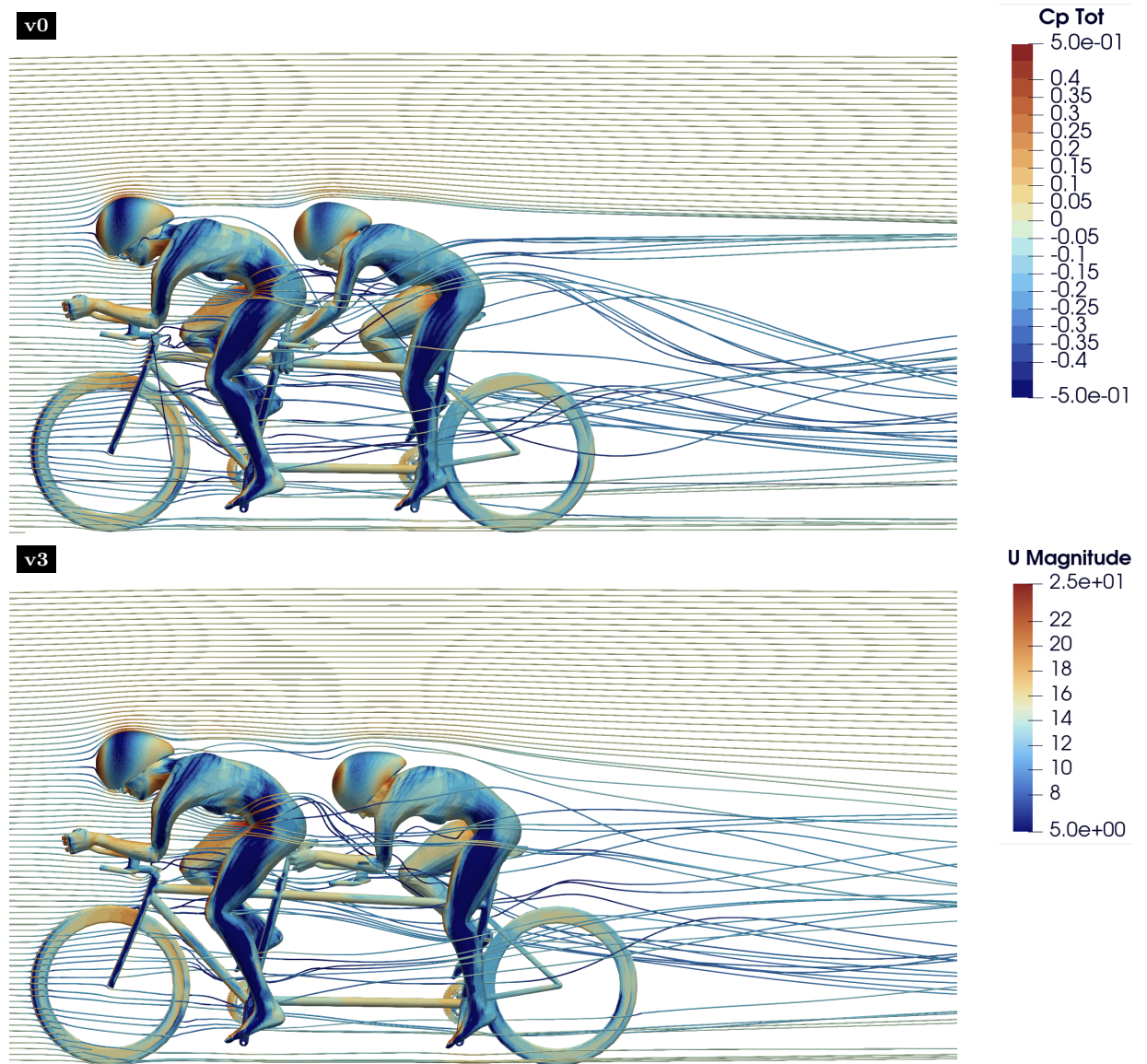


Figure 5.6: Streamline visualization for configurations v0 (top) and v3 (bottom). The riders' surfaces are coloured by total pressure coefficient ($C_{p,tot}$), while the streamlines are coloured by velocity magnitude [m/s].

5.6. Combined Configurations and Non-Additive Effects

Combined configurations demonstrate that aerodynamic improvements are not strictly additive. Configuration v5, which combines pilot head lowering with reduced inter-rider spacing, yields the lowest global C_dA among tandem cases. In this configuration, the intrinsic reduction in pilot drag due to posture optimisation is reinforced by favourable wake interaction and pressure recovery effects. The wake development representation is shown in Figure 5.7.

In contrast, configuration v6 combines pilot head lowering with a rearward stoker position. Despite the improved pilot posture, the pilot drag in v6 exceeds that of both the baseline configuration v0 and the pilot-optimised configuration v2. This result highlights the dominant role of interaction effects: the increased inter-rider spacing allows the pilot wake to develop more freely, reducing base pressure on the pilot rear surface and introducing an interaction-driven penalty that, in tandem configuration, outweighs the intrinsic benefit of head lowering.

These findings demonstrate that aerodynamic optimisation in tandem configurations requires careful balancing of posture improvements and interaction effects, as modifications beneficial to one rider may penalise the other.

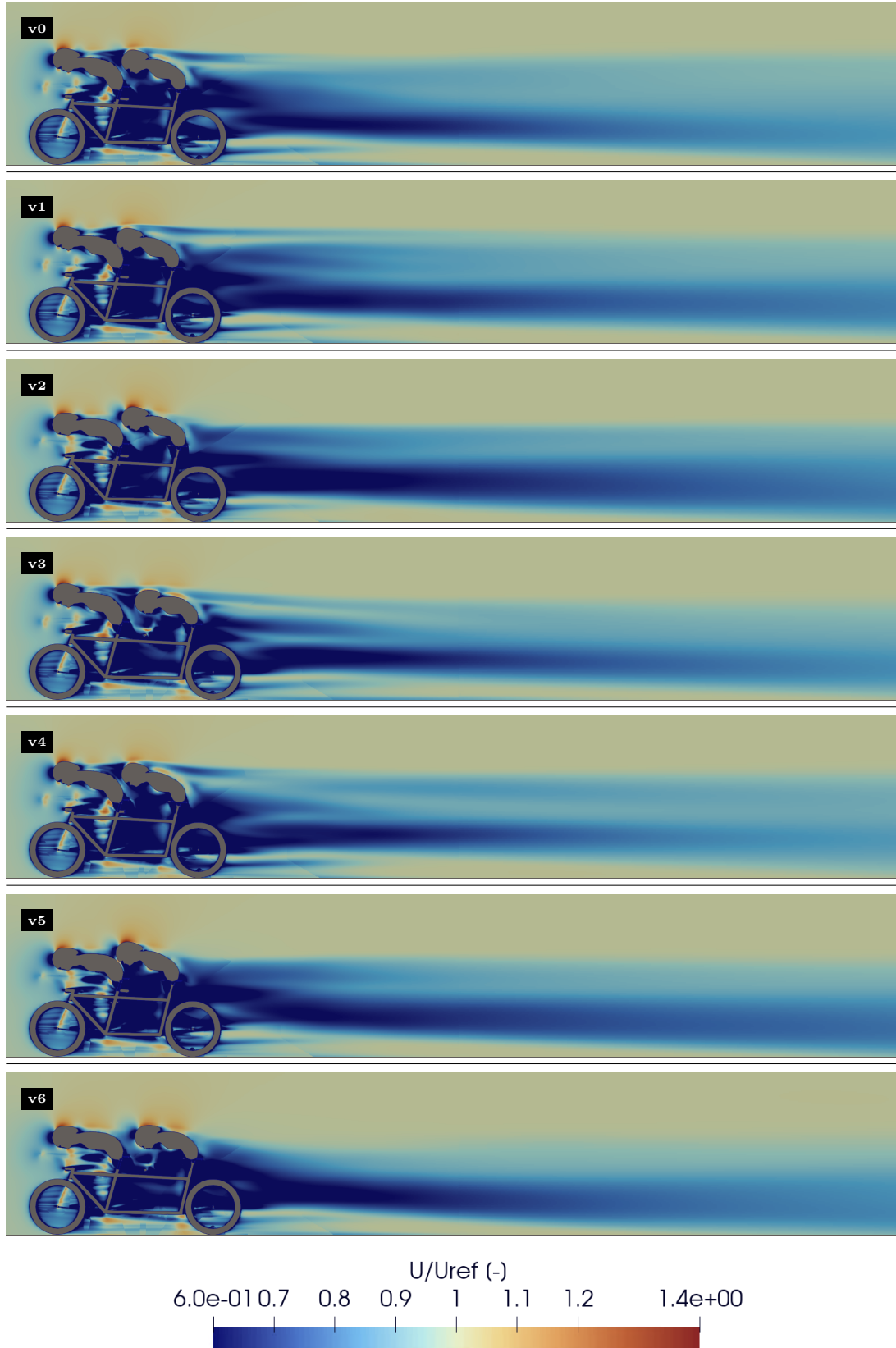


Figure 5.7: Vertical mid-plane slice of the normalized velocity magnitude (U/U_{ref}) for configurations v0–v6, illustrating the wake development downstream of the tandem system. The slice is extracted along the longitudinal mid-plane of the computational domain.

5.7. Single-Rider Reference Cases

Pilot-only configurations v7 and v8 provide a baseline to isolate the intrinsic aerodynamic effect of pilot posture. The comparison in Figure 5.8 confirms that lowering the pilot head significantly reduces drag in isolation. When the stoker is introduced, the pilot drag is further modified by wake interaction effects, which can either amplify or offset the intrinsic posture-related benefit depending on inter-rider spacing.

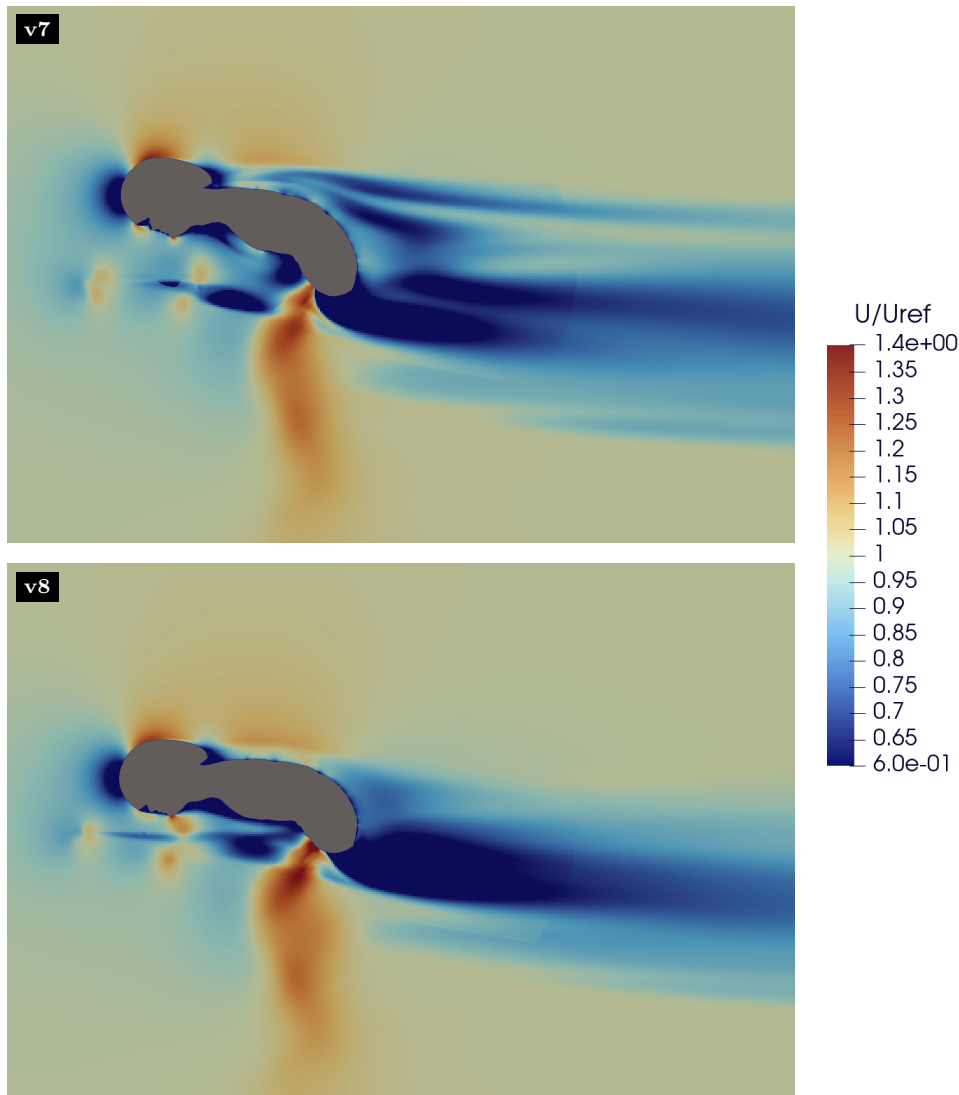


Figure 5.8: Vertical mid-plane slice of the normalized velocity magnitude (U/U_{ref}) for the pilot-only configurations. The plane corresponds to the longitudinal mid-plane of the domain and highlights the flow acceleration over the rider and the wake development downstream.

5.8. Cumulative Drag Distributions Along Streamwise and Vertical Directions

Cumulative drag distributions along the streamwise and vertical directions in Figures 5.9 and 5.10 provide a spatial interpretation of where aerodynamic loads are generated. The analysis indicates that most drag differences originate from the upper-body region ($y > 0.9$ m) and from the rear sections of both riders ($x > 0.7$ m and $x > 1.6$ m), corresponding to the helmet, shoulders and wake closure regions.

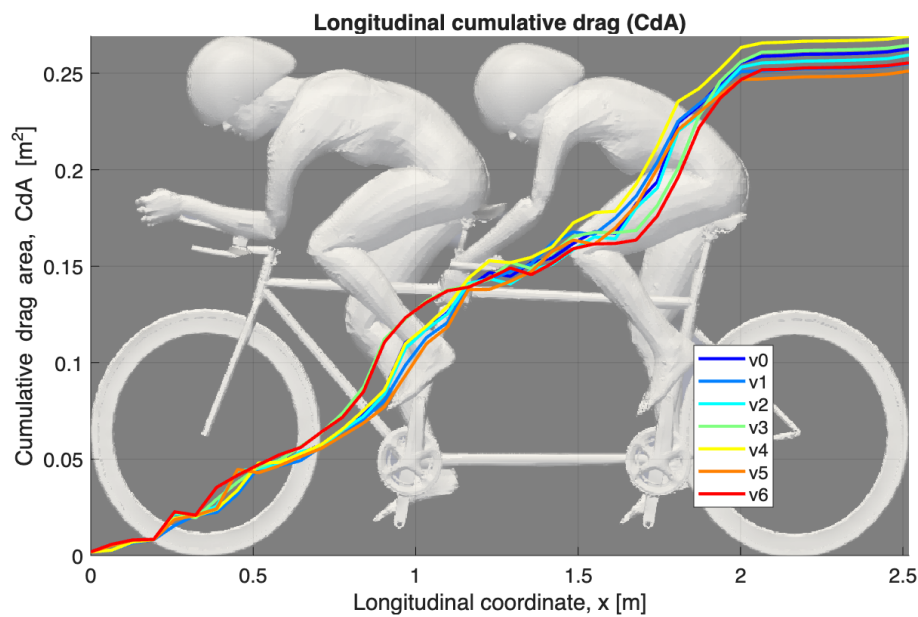


Figure 5.9: Longitudinal cumulative drag distribution along the streamwise direction for the analysed tandem configurations. The curves represent the cumulative drag area contribution (C_dA) obtained by integrating the pressure and viscous forces over the entire external surface of the tandem system (bike frame, pilot, and stoker) from the most upstream point to the considered streamwise position.

These results confirm that relatively small geometric changes in these areas can produce measurable variations in total drag, particularly in tandem configurations where wake interaction effects dominate.

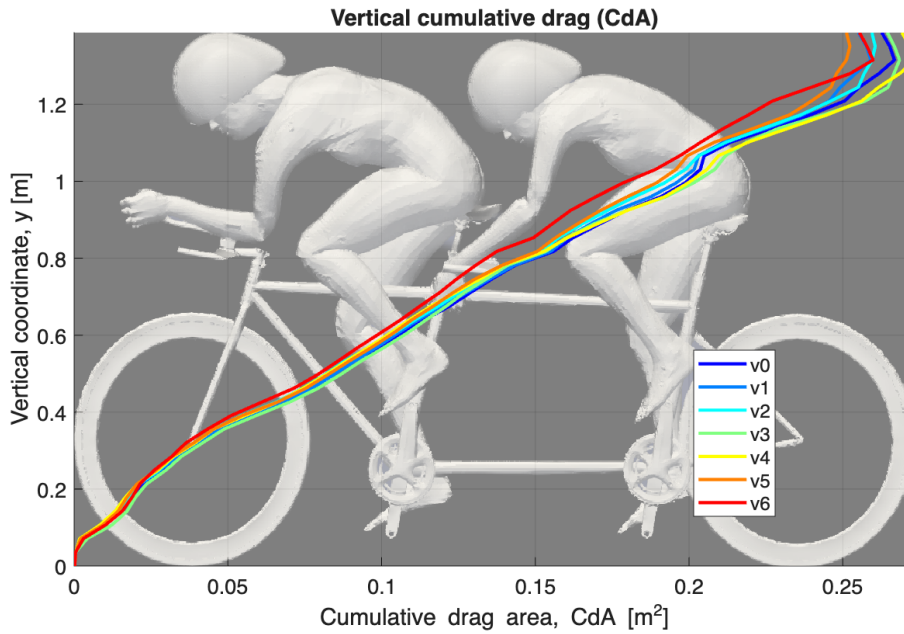


Figure 5.10: Vertical cumulative drag distribution as a function of the vertical coordinate for the analysed tandem configurations.

5.9. Summary of Aerodynamic Findings

The main aerodynamic findings of this chapter can be summarised as follows:

- Pilot head alignment is the most effective single modification for reducing intrinsic pilot drag.
- Inter-rider spacing strongly controls aerodynamic interaction effects and can either reduce or increase pilot drag depending on the resulting wake dynamics.
- Stoker posture optimisation can reduce stoker drag even under reduced shielding conditions.
- Aerodynamic improvements in tandem configurations are non-additive due to non-linear wake coupling.
- The best-performing configuration (v5) combines pilot posture optimisation with controlled stoker proximity, achieving a favourable balance between intrinsic and interaction-driven effects.

6 | Biomechanical Results

This chapter presents the biomechanical results obtained from the experimental campaign and subsequent data processing. The analysis integrates the outputs derived from the musculoskeletal model with experimentally measured electromyography (EMG) signals, with the aim of characterising posture-dependent adaptations, muscle coordination strategies, and fatigue development in tandem cycling.

6.1. Kinematic Results

6.1.1. Overview of Kinematic Metrics

Kinematic results were obtained from the musculoskeletal model using full-body motion data acquired through inertial measurement units, as reported in Chapter 4.

Since cycling kinematics are expected to remain unchanged across different power outputs, the present analysis is based exclusively on the 5-minute trial performed at 300 W and 100 rpm. This condition was selected as it minimises transient effects associated with short-duration efforts and provides a stable description of posture and joint motion during sustained cycling.

Reported kinematic metrics include mean postural angles and their standard deviations over time, as well as minimum, maximum, and range of motion (ROM) values for the hip, knee, and ankle joints of both the right and left legs.

Body angles have been defined according to the geometric conventions illustrated in Figure 6.1. The reported angles are measured in the sagittal plane and are referenced to anatomically meaningful segments and axes. For clarity, the figures show representative angle values corresponding to the road configuration (right side) as illustrative examples, while the same conventions are consistently applied to all riding positions and both limbs throughout the analysis.

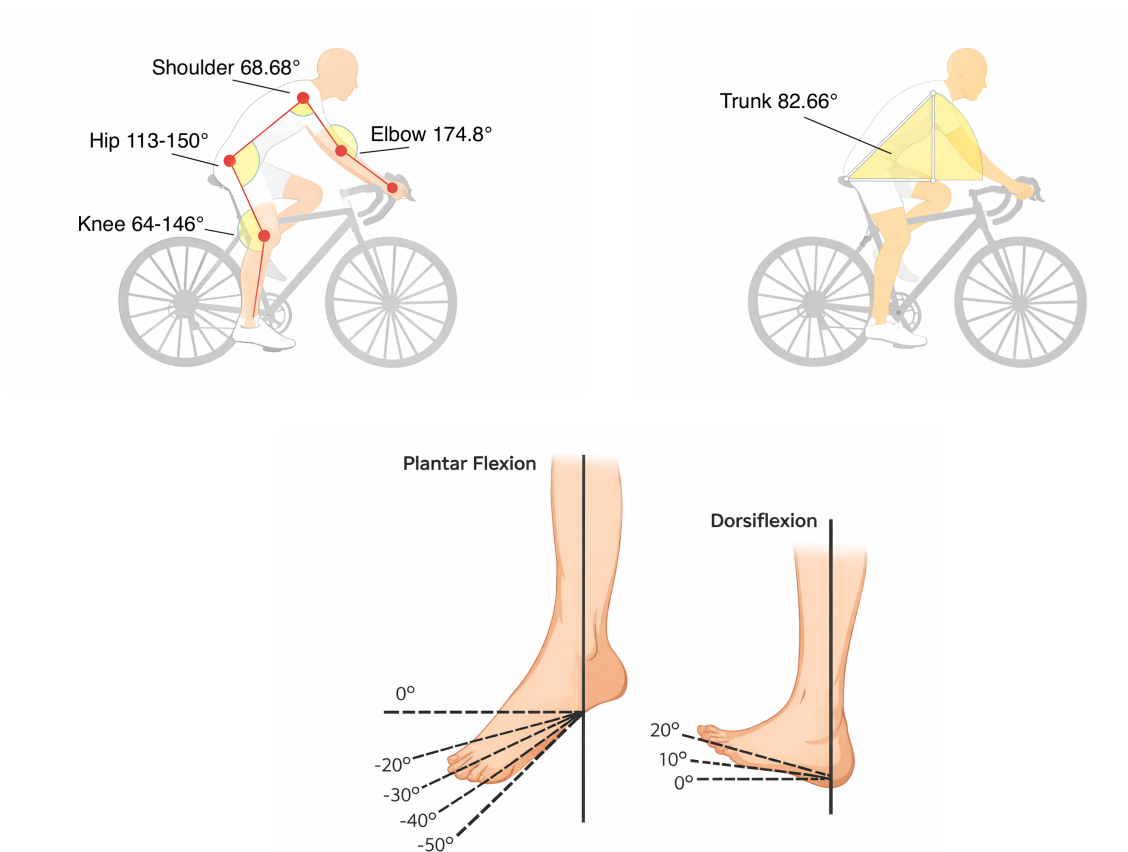


Figure 6.1: Geometric conventions adopted for the definition of kinematic angles in the sagittal plane, including the definition of upper- and lower-limb joint angles, the trunk angle with respect to the horizontal reference, and the ankle plantarflexion and dorsiflexion angle convention. Representative values from the road configuration (right side) are shown for illustrative purposes only.

6.1.2. Upper-Body Posture: Trunk and Upper Limbs

Upper-body posture was characterised by analysing the trunk inclination angle relative to the horizontal plane, together with shoulder and elbow joint angles. Mean values and standard deviations over time are reported in Table 6.1.

A progressive reduction in trunk inclination is observed when moving from the road position to the frame-clench and time-trial configurations. The road position exhibits the most upright posture, whereas the frame-clench configuration shows the lowest trunk angle, indicating a pronounced forward lean. The time-trial position presents an intermediate trunk inclination, reflecting the combined effect of aerodynamic requirements and forearm support on aerobars.

Upper-limb joint angles show marked posture-dependent differences. Compared to the road position, both the frame-clench and time-trial configurations are characterised by increased shoulder angle and reduced elbow angle. In particular, the time-trial position exhibits a substantial increase in elbow flexion, consistent with sustained forearm support on the aerobars.

For all upper-body angles, the standard deviation over time remains limited, indicating a stable posture throughout the pedalling cycle.

Table 6.1: Mean upper-body postural angles (mean $\pm$ standard deviation) for the road, frame-clench, and time-trial positions during the 5-minute steady-state trial at 300 W and 100 rpm.

Angle	Road	Frame-clench	Time-trial
Trunk [°]	82.66 $\pm$ 1.15	71.17 $\pm$ 1.17	73.90 $\pm$ 1.24
Shoulder [°]	68.68 $\pm$ 1.85	84.42 $\pm$ 2.36	93.39 $\pm$ 1.59
Elbow [°]	174.80 $\pm$ 1.32	143.29 $\pm$ 2.42	104.23 $\pm$ 1.33

6.1.3. Lower-Limb Joint Kinematics and Range of Motion

Lower-limb kinematics were analysed for the hip, knee, and ankle joints of both the right and left legs. Minimum and maximum joint angles over the pedalling cycle were extracted, and the corresponding range of motion (ROM) was computed.

Table 6.2 reports joint ROM values for all riding positions. Across all configurations, ROM values fall within physiologically plausible ranges for cycling.

Posture-dependent differences are most evident at the hip and ankle joints. Compared

to the road position, both the frame-clench and time-trial configurations are associated with reduced hip joint angles, reflecting the more flexed hip posture required by increased trunk inclination.

Ankle joint ROM shows moderate variations between riding positions, suggesting compensatory adaptations to preserve effective power transmission despite changes in torso orientation and upper-body support.

In contrast, knee joint ROM exhibits relatively small variations across riding positions. This relative invariance indicates that knee kinematics are primarily constrained by crank geometry and saddle height rather than by upper-body posture.

Bilateral comparisons reveal similar ROM values between the right and left legs for all joints, indicating no pronounced asymmetries under the controlled experimental conditions.

Table 6.2: Lower-limb joint range of motion (ROM) across riding positions. Values are reported as minimum–maximum angle and mean ROM $\pm$ standard deviation (in degrees).

Joint	Side	Road	FC	TT
Hip	R	113.29–150.27	99.40–138.85	103.27–140.39
		36.98 $\pm$ 0.25	39.45 $\pm$ 0.32	37.13 $\pm$ 0.86
	L	109.01–150.08	99.99–142.56	103.50–145.63
		41.07 $\pm$ 0.49	42.57 $\pm$ 0.57	42.13 $\pm$ 1.20
Knee	R	64.55–146.63	65.03–149.21	59.81–138.19
		82.09 $\pm$ 0.80	84.18 $\pm$ 0.75	78.38 $\pm$ 0.89
	L	67.87–143.93	65.70–148.67	65.19–138.55
		76.06 $\pm$ 0.75	82.96 $\pm$ 0.72	73.37 $\pm$ 0.10
Ankle	R	-50.67–7.80	-47.56–13.79	-45.56–11.81
		58.46 $\pm$ 1.87	61.36 $\pm$ 1.87	57.37 $\pm$ 2.01
	L	-41.03–18.76	-44.38–16.32	-41.05–16.66
		59.78 $\pm$ 1.13	60.70 $\pm$ 1.13	57.71 $\pm$ 1.74

6.1.4. Kinematic Consistency and Modelling Considerations

Overall, the kinematic results demonstrate that the musculoskeletal model reproduces stable and posture-dependent joint kinematics across all investigated riding configurations. Observed trends in trunk inclination, upper-limb posture, and lower-limb joint motion align with expected biomechanical adaptations to increasingly aerodynamic cycling positions.

6.2. EMG Activation Patterns

This section presents the analysis of cycle-averaged muscle activation patterns derived from surface electromyography (EMG), as described in Chapter 4.

6.2.1. Effect of Power Output and Cadence

Within each riding position, EMG activation patterns were compared across the four experimental tests to assess the influence of power output and cadence on the temporal structure of muscle activation. The results are reported in Figures 6.2, Figure 6.3 and Figure 6.4.

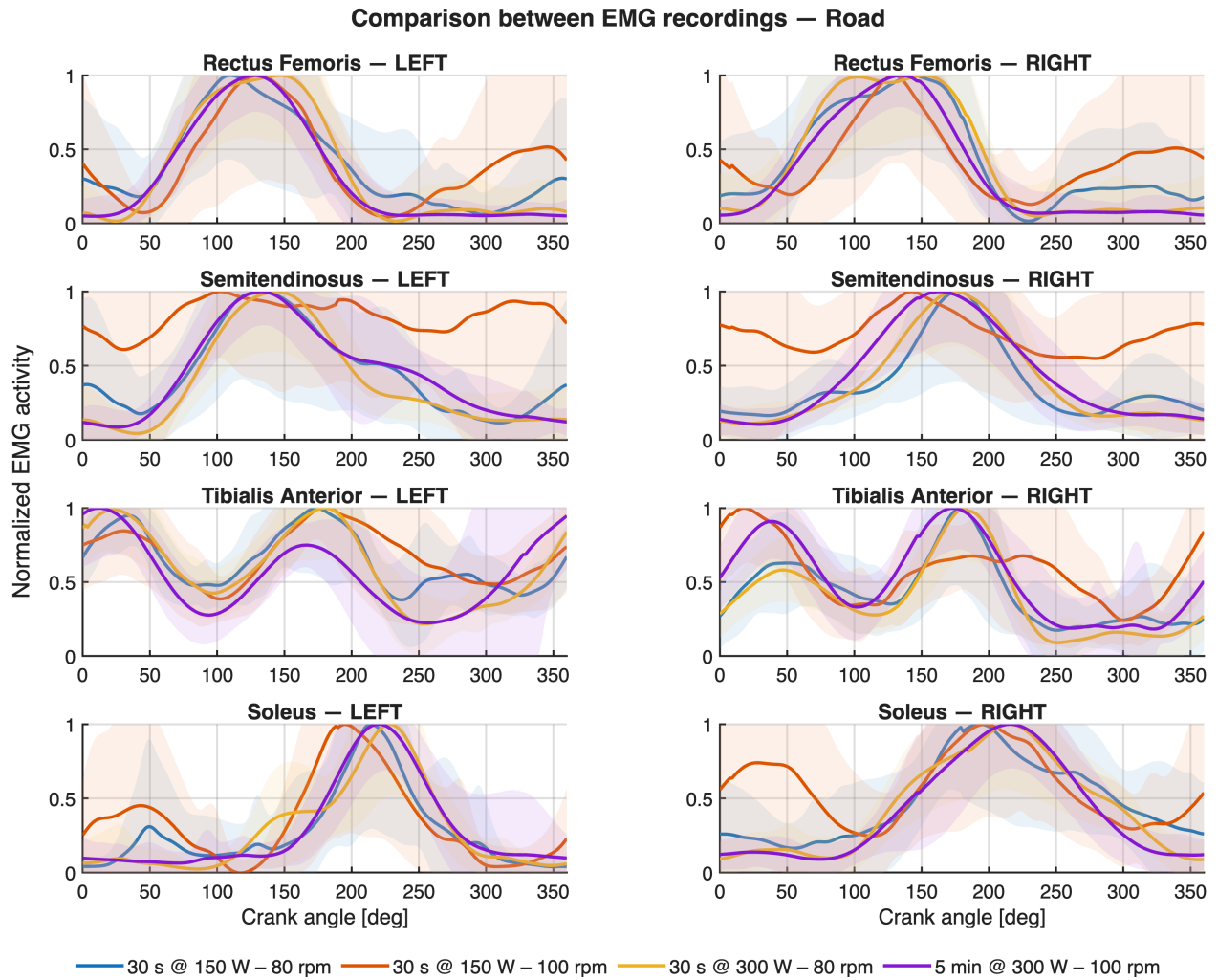


Figure 6.2: EMG envelope comparison for the road configuration.

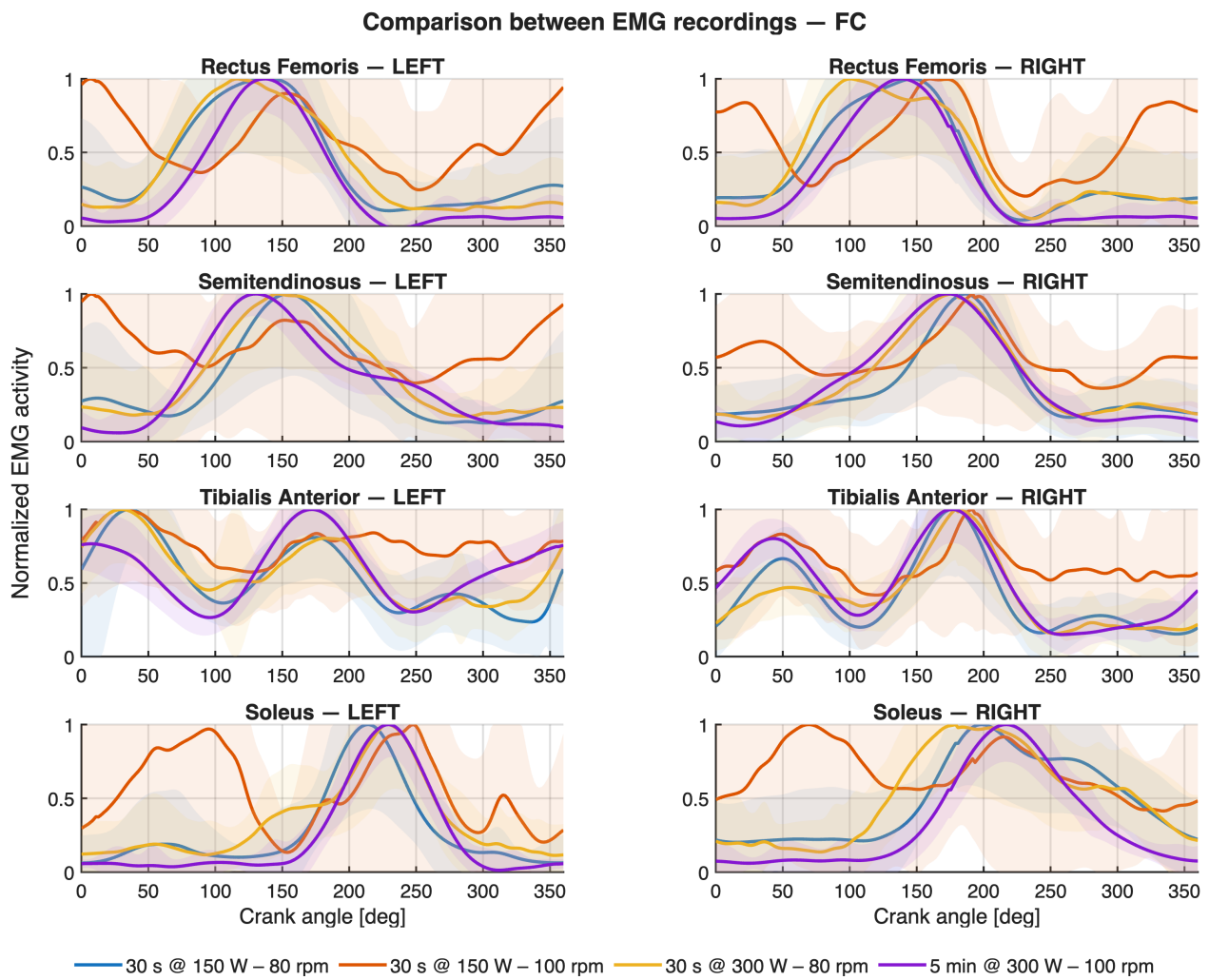


Figure 6.3: EMG envelope comparison for the frame-clench configuration.

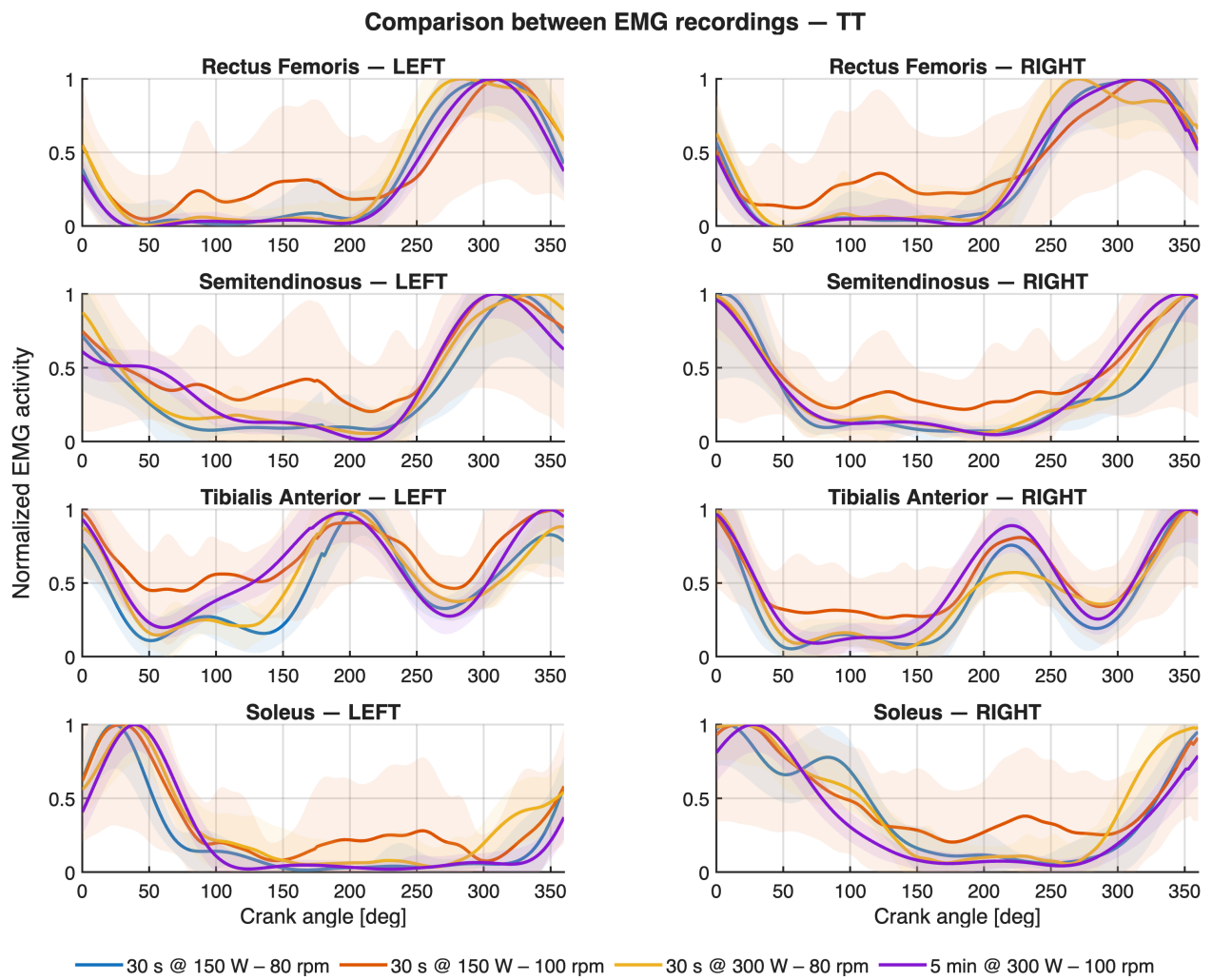


Figure 6.4: EMG envelope comparison for the time-trial configuration.

Across all muscles and positions, the overall shape and timing of EMG activation remain highly consistent between tests. No systematic shifts in activation onset, offset, or peak timing are observed when varying either power output or cadence, indicating a robust neuromuscular coordination strategy across the explored conditions.

An exception is observed for the 30 s trial performed at 150 W and 100 rpm, which exhibits noticeably increased signal variability and reduced smoothness in the cycle-averaged EMG profiles. This effect is consistently observed across multiple muscles and riding positions and is probably attributed to experimental factors associated with the specific test condition. In particular, the combination of very low power output and unusually high cadence likely resulted in reduced mechanical stability, with the rider pedalling close to unloaded conditions and exhibiting increased vertical oscillations on the saddle. Such conditions are expected to amplify motion-related artefacts and soft-tissue vibrations, leading to noisier EMG recordings.

6.2.2. Effect of Riding Position

To investigate posture-dependent effects on neuromuscular coordination, EMG activation patterns were compared across the road, frame-clench, and time-trial positions under identical power and cadence conditions (300 W, 100 rpm) during 5-minute trials. The results of these comparisons are shown in Figure 6.5.

Across all analysed muscles, activation timing relative to crank angle remains highly consistent between riding positions. No systematic differences in activation onset, duration, or phase are observed when transitioning from the road to more aerodynamic configurations.

Given the normalisation of EMG amplitude, the absence of clear posture-dependent differences indicates that the fundamental coordination pattern of the pedalling task is preserved across riding positions. This suggests that changes in upper-body posture and support conditions do not substantially alter the timing structure of lower-limb muscle activation during cycling.

6.2.3. Muscle-Specific Considerations

Despite differences in functional role, all analysed muscles exhibit activation patterns that are consistent with established descriptions of cycling biomechanics [5]. Tibialis anterior is primarily active during the recovery phase, soleus during the downstroke, and rectus femoris and semitendinosus show broader activation windows spanning phase transitions.

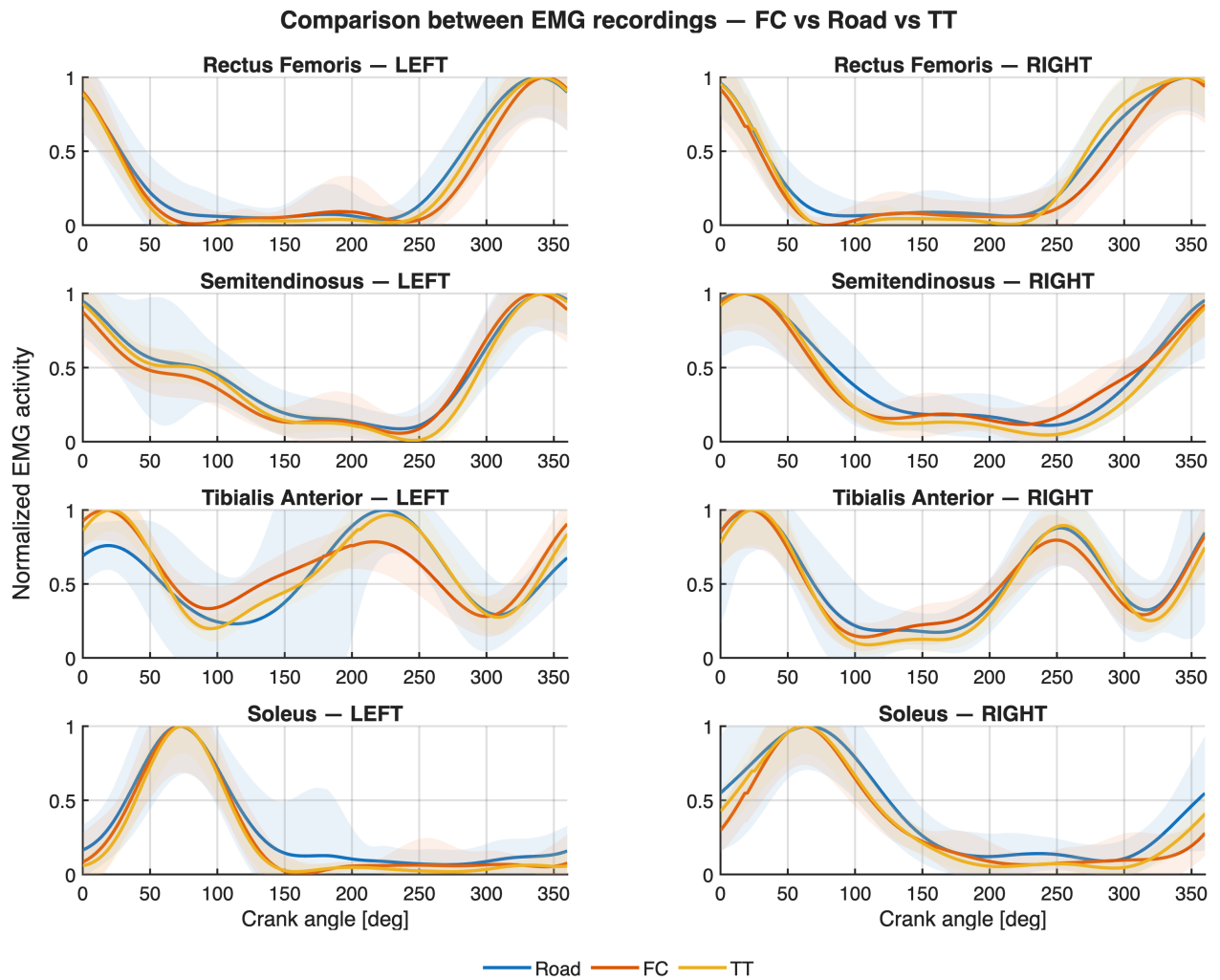


Figure 6.5: Comparison of cycle-averaged EMG activation patterns between road, frame-clench, and time-trial positions under identical mechanical conditions. EMG amplitude is normalised.

Across muscles, the preservation of activation timing across power, cadence, and posture conditions supports the interpretation that cycling neuromuscular coordination is highly stereotyped under steady pedalling conditions.

6.2.4. Summary of EMG Activation Findings

In summary, cycle-averaged EMG activation patterns demonstrate a high degree of robustness across riding positions, power outputs, and cadences. When EMG amplitude is normalised, no systematic differences in activation timing or waveform shape are observed between conditions.

The only notable deviation is the increased signal variability observed during the 30 s trial at 150 W and 100 rpm, which is attributed to reduced mechanical stability and increased motion artefacts rather than to changes in neuromuscular activation.

These findings justify the subsequent focus on the 5-minute steady-state trial for EMG–model comparisons and fatigue analysis, where both mechanical and neuromuscular conditions are more controlled and physiologically meaningful.

6.3. Comparison Between EMG Measurements and Musculoskeletal Model Predictions

This section compares experimentally measured EMG activation patterns with muscle activation estimates obtained from the musculoskeletal model. The comparison is limited to the 5-minute steady-state cycling trial performed at 300 W and 100 rpm, for the previously stated reason.

The objective of this analysis is not to validate the musculoskeletal model quantitatively, but rather to assess whether inverse-dynamics-based muscle recruitment is able to reproduce the main qualitative features of cycling muscle coordination and to identify systematic discrepancies between model predictions and experimental measurements.

6.3.1. Methodological Scope of the Comparison

Muscle activation estimates were obtained from the AnyBody inverse dynamics analysis using the experimentally measured kinematics and external constraints. Predicted activations were phase-normalised with respect to crank angle and compared to cycle-averaged EMG activation profiles.

Both EMG and model-based activations were normalised, allowing qualitative comparison of activation timing and waveform shape while avoiding any interpretation in terms of absolute activation magnitude. This choice is essential, as surface EMG amplitude and model-predicted muscle activation are not directly comparable quantities.

The comparison focuses on four representative muscles: tibialis anterior, soleus, rectus femoris, and semitendinosus, and is performed separately for the road, frame-clench, and time-trial positions.

6.3.2. Qualitative Agreement and Discrepancies

Figures 6.6–6.8 report the cycle-averaged EMG activation profiles alongside the corresponding muscle activations predicted by the musculoskeletal model for the road, frame-clench, and time-trial configurations.

For mono-articular muscles with a well-defined mechanical role, such as the soleus, partial qualitative agreement is observed. In these cases, the model captures the general phase of activation during the downstroke, consistent with the muscle’s contribution to ankle plantarflexion and power transmission.

In contrast, for bi-articular muscles such as the rectus femoris and semitendinosus, substantial discrepancies are observed between experimentally measured EMG signals and model-predicted activations. Differences include shifts in activation timing, altered activation duration, and, in some cases, activation peaks occurring in phases where EMG activity is minimal.

The tibialis anterior shows limited correspondence between EMG and model predictions, with the model often producing narrow activation bursts that do not reflect the broader and more distributed activation observed experimentally during the recovery phase.

These discrepancies are consistently observed across all riding positions and do not appear to be posture-specific.

6.3.3. Interpretation of Model–EMG Mismatch

The observed discrepancies between EMG measurements and musculoskeletal model predictions reflect well-known limitations of inverse-dynamics-based muscle recruitment in cycling applications. The muscle redundancy problem is inherently underdetermined, and the resulting activation patterns depend strongly on the selected optimisation criterion rather than on subject-specific neuromuscular control strategies.

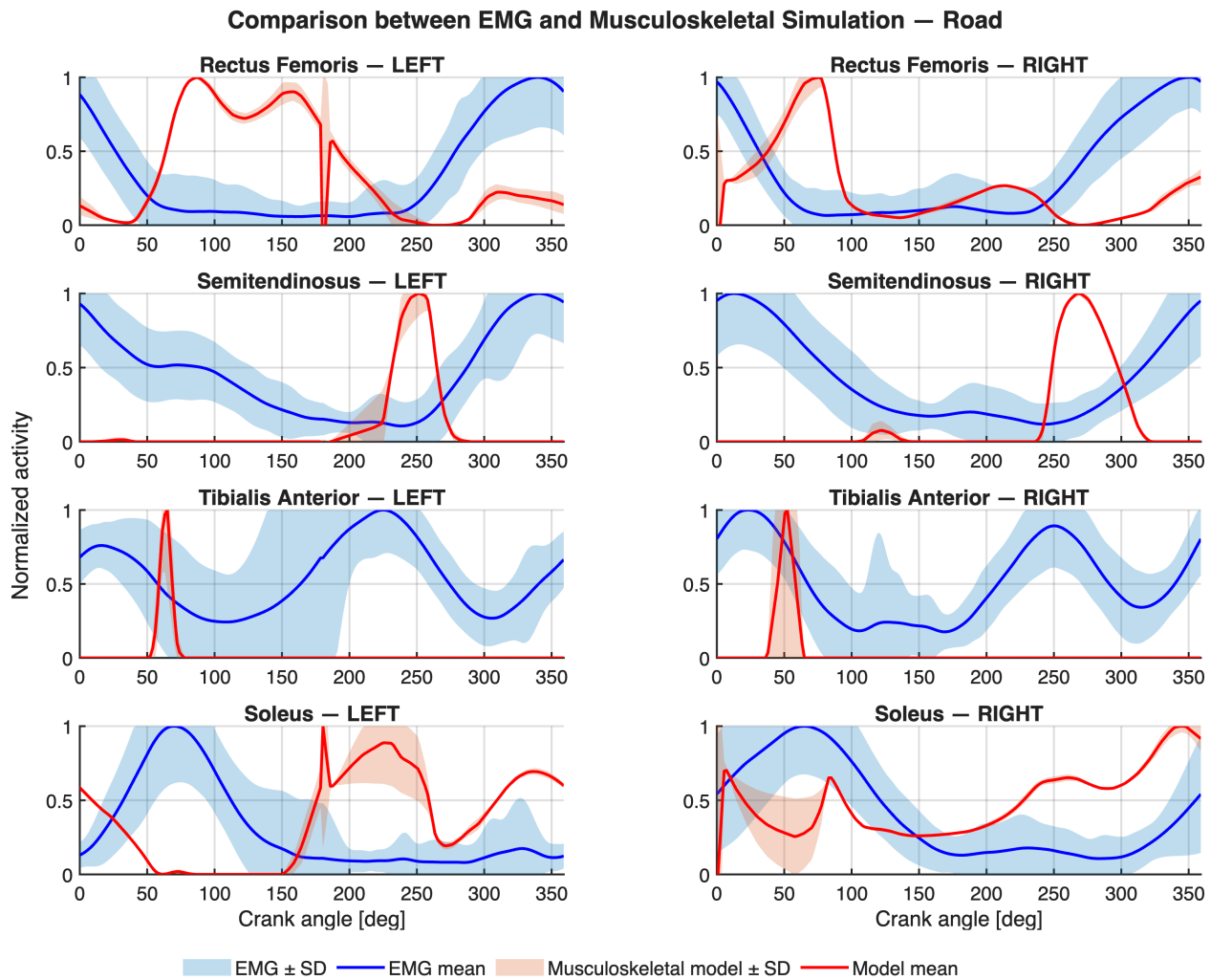


Figure 6.6: Comparison between cycle-averaged EMG activation profiles and musculoskeletal model predictions for the road configuration during the 5-minute trial at 300 W and 100 rpm.

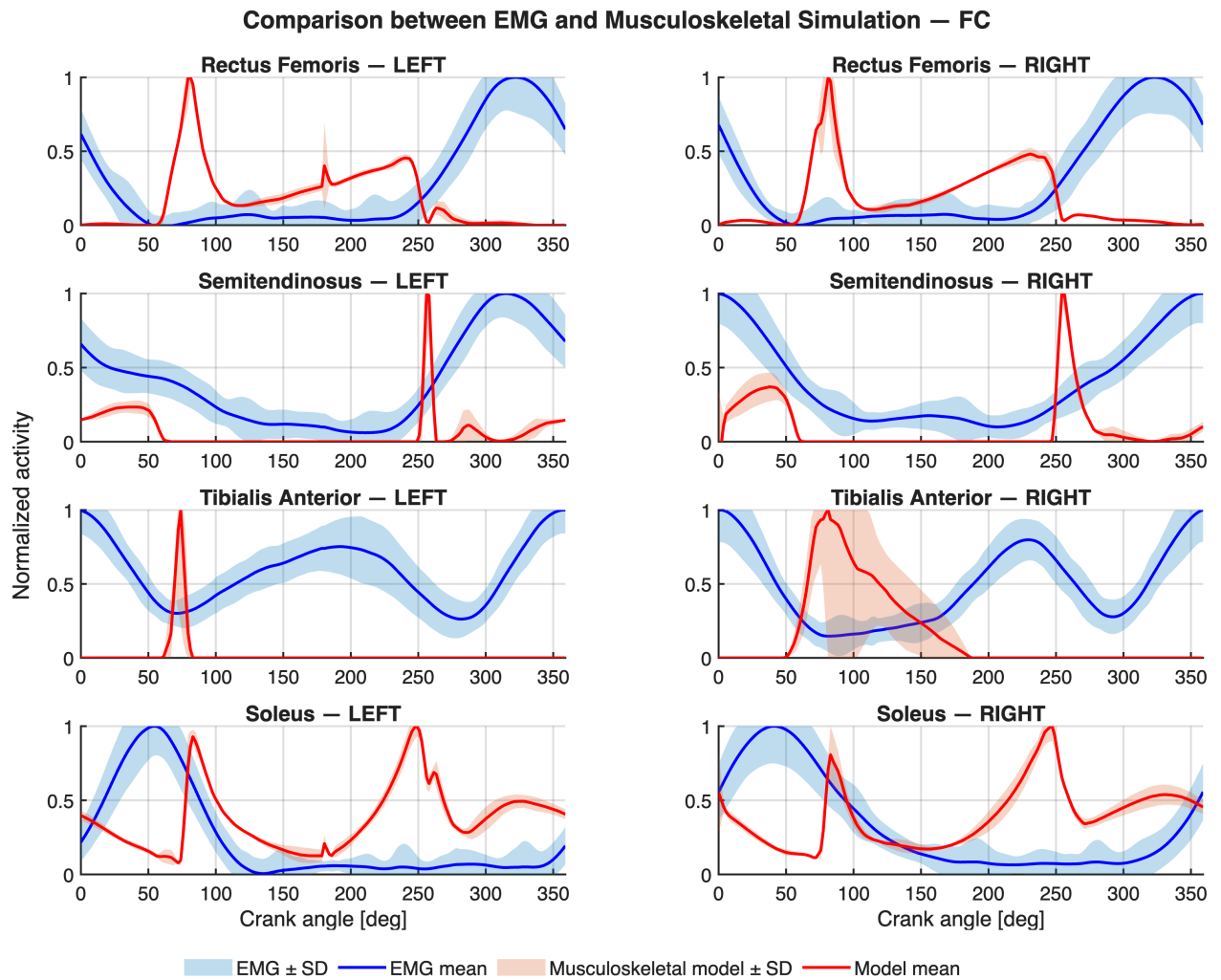


Figure 6.7: Comparison between cycle-averaged EMG activation profiles and musculoskeletal model predictions for the frame-clench configuration during the 5-minute trial at 300 W and 100 rpm.

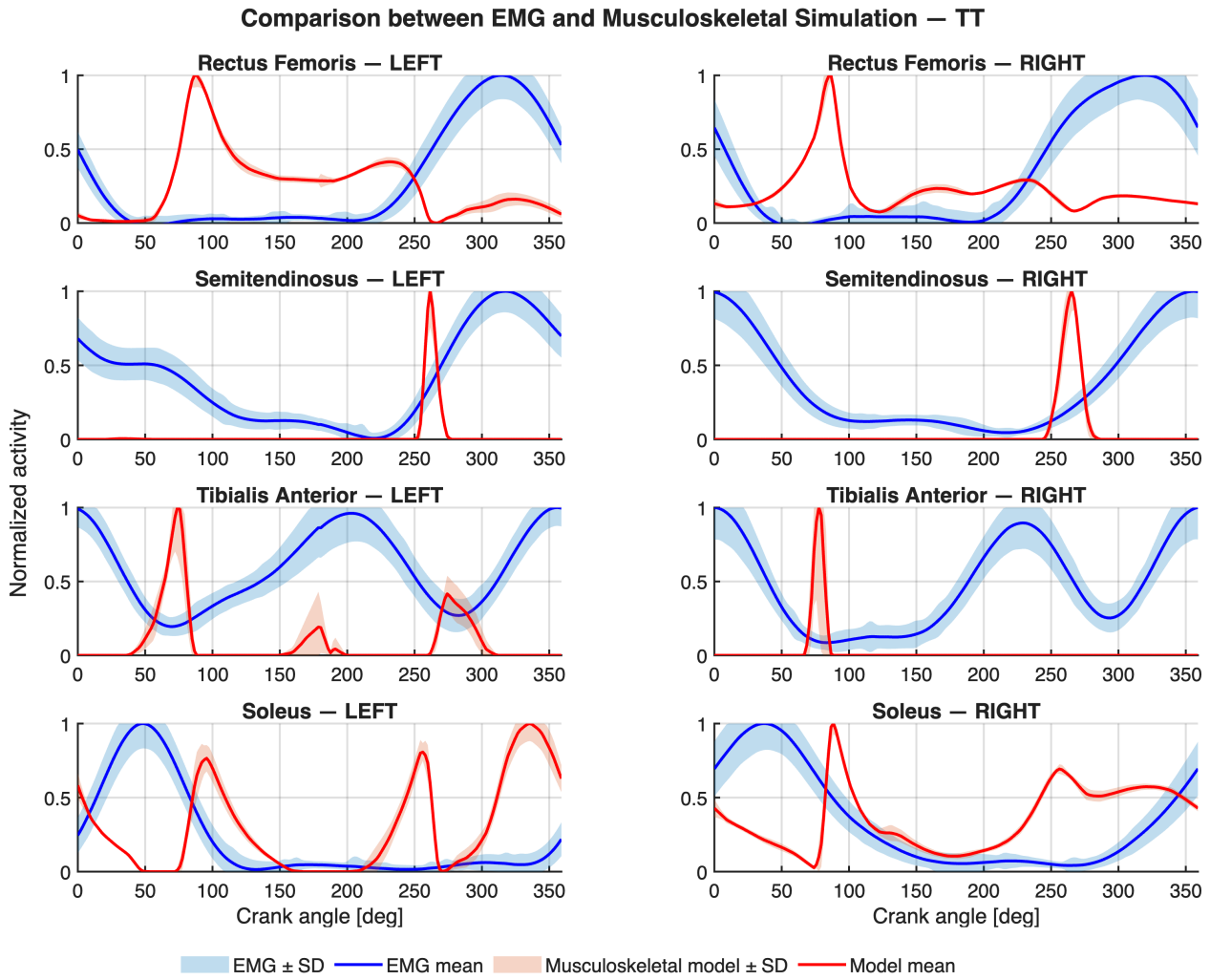


Figure 6.8: Comparison between cycle-averaged EMG activation profiles and musculoskeletal model predictions for the time-trial configuration during the 5-minute trial at 300 W and 100 rpm.

In real cycling, muscle coordination is influenced by factors such as fatigue management, joint stabilisation and learned motor patterns, none of which are explicitly represented in the model. As a result, agreement between model-predicted muscle activations and EMG measurements is expected to be qualitative at best, particularly for bi-articular muscles whose function cannot be inferred unambiguously from joint-level torques alone.

6.3.4. Antagonistic Muscle Behaviour

Despite the discrepancies observed between experimentally measured EMG signals and musculoskeletal model predictions at the level of individual muscles, the model successfully reproduces a key qualitative feature of cycling biomechanics: antagonistic muscle behaviour. When the predicted activations are considered jointly, functionally antagonistic muscle groups exhibit complementary activation patterns over the pedalling cycle.

At the knee and hip joints, the bi-articular muscles rectus femoris and semitendinosus act as functional antagonists. Rectus femoris contributes primarily to knee extension and hip flexion, while semitendinosus is associated with knee flexion and hip extension. In the musculoskeletal model predictions, these muscles exhibit activation phases that are largely non-overlapping, indicating that joint moments are generated through mechanically consistent agonist–antagonist coordination.

Importantly, this antagonistic structure is preserved across the road, frame-clench, and time-trial positions. Although the precise timing and duration of individual muscle activations do not always align with EMG measurements, the absence of spurious simultaneous activation of antagonist muscles suggests that the optimisation-based recruitment strategy enforces physiologically meaningful joint-level behaviour.

This result is non-trivial in the context of cycling biomechanics, where muscle redundancy and the cyclic nature of the task could otherwise lead to unrealistic co-activation patterns. The correct qualitative separation between agonist and antagonist muscle groups supports the mechanical plausibility of the predicted joint torques.

6.3.5. Implications for the Present Study

Based on the results of this comparison, musculoskeletal modelling was not employed as a quantitative predictor of muscle activation or neuromuscular fatigue in the present study. Experimental EMG measurements were therefore adopted as the primary reference for analysing muscle activation patterns and fatigue development.

The musculoskeletal model was retained exclusively for kinematic processing, includ-

ing crank-angle computation, cycle segmentation, and phase normalisation of EMG signals. This separation ensures a clear methodological distinction between experimentally validated measurements and model-based kinematic support, while avoiding over-interpretation of model-derived muscle activation estimates.

6.4. Fatigue Analysis

This section presents the analysis of neuromuscular fatigue derived from surface EMG measurements. Fatigue-related indicators are evaluated exclusively for the 5-minute steady-state cycling trial performed at 300 W and 100 rpm, which represents the most demanding and physiologically meaningful condition of the experimental protocol.

Fatigue assessment is based on temporal changes in EMG signal characteristics. Frequency-domain indicators, namely median frequency (MF) and mean frequency (MNF), are used as primary markers of neuromuscular fatigue, while the root mean square (RMS) amplitude is analysed as a complementary indicator reflecting changes in muscle activation and potential compensatory recruitment strategies. In classical interpretations, negative MF and MNF trends are associated with a spectral shift towards lower frequencies and are commonly linked to the development of neuromuscular fatigue. However, the magnitude and consistency of these trends must be interpreted in relation to the intrinsic variability of EMG signals and to possible non-monotonic behaviour over time.

6.4.1. Time-Based Fatigue Indicators

For each muscle and riding position, robust linear slopes of MF, MNF, and RMS were computed over the entire duration of the trial. In addition, relative percentage changes between early and late phases of the trial were evaluated to characterise phase-dependent differences in activation patterns. The results are reported in Table 6.3.

The statistical analysis reported in Table 6.4 provides formal support for the temporal trends described in Table 6.3. The regression-based p_{slope} assesses whether the MF evolution differs significantly from a null (zero-slope) hypothesis, whereas the paired Wilcoxon early-late test ($p_{\text{E-L}}$) evaluates whether the second half of the trial differs systematically from the first. The corresponding ΔMF values quantify the magnitude of these early-late spectral shifts and therefore provide complementary information on effect size.

In the road configuration, statistically significant negative MF slopes are observed bilaterally for the tibialis anterior ($p < 10^{-7}$), accompanied by early-late reductions of 8–10%. Additional significant trends are detected for selected distal muscles (e.g., SOL-R

Table 6.3: Time-based EMG fatigue indicators (bilateral) for the 5-minute trial at 300 W and 100 rpm. Percentage variations refer to early–late comparison over the trial duration.

Position	Muscle	MF slope	MNF slope	RMS slope	Δ MF [%]	Δ MNF [%]	Δ RMS [%]
Road	TA-R	-0.1525	-0.1606	1.63×10^{-5}	-8.36	-7.11	3.75
	TA-L	-0.1723	-0.1598	-5.62×10^{-5}	-9.65	-7.38	-7.72
	ST-R	0.0032	-0.0093	2.97×10^{-4}	-1.15	-1.63	44.45
	ST-L	-0.1600	-0.1061	1.42×10^{-4}	-7.47	-3.58	14.70
	RF-R	-0.0150	0.0062	1.66×10^{-4}	1.35	2.13	10.24
	RF-L	-0.0710	-0.0796	2.80×10^{-4}	0.88	-0.94	8.35
	SOL-R	-0.0421	-0.0528	8.49×10^{-5}	-5.15	-4.65	15.68
	SOL-L	-0.0492	-0.0339	-1.59×10^{-4}	-2.74	-1.62	-9.72
Frame-clench	TA-R	-0.0846	-0.0775	-8.87×10^{-5}	-3.33	-2.66	-11.53
	TA-L	0.0200	0.0207	-3.60×10^{-5}	1.20	1.18	-6.18
	ST-R	0.0075	0.0290	1.21×10^{-4}	0.69	1.17	15.55
	ST-L	-0.0142	-0.0156	5.93×10^{-5}	-0.05	-0.14	7.54
	RF-R	0.0406	0.0628	-5.04×10^{-5}	3.83	4.10	-1.96
	RF-L	-0.1422	-0.1619	3.21×10^{-4}	-7.82	-8.40	15.54
	SOL-R	-0.0503	-0.0626	2.79×10^{-5}	-2.88	-2.56	2.70
	SOL-L	0.0023	-0.0106	3.01×10^{-5}	0.45	-0.04	3.59
Time-trial	TA-R	0.0190	0.0230	-1.87×10^{-4}	1.67	1.48	-19.26
	TA-L	-0.0190	-0.0302	1.11×10^{-5}	0.39	-0.05	0.87
	ST-R	0.0141	-0.0011	9.81×10^{-5}	0.60	-0.38	14.00
	ST-L	-0.0614	-0.0228	-9.61×10^{-6}	-3.58	-1.77	-0.01
	RF-R	-0.0795	-0.0641	8.84×10^{-5}	-5.21	-3.34	4.55
	RF-L	-0.0803	-0.0974	1.79×10^{-4}	-4.47	-5.17	8.12
	SOL-R	-0.0240	-0.0235	-3.64×10^{-5}	-1.88	-1.34	-1.79
	SOL-L	-0.0239	-0.0159	-9.91×10^{-5}	-0.29	-0.94	-2.60

Table 6.4: Statistical significance of time-based MF trends. p_{slope} refers to the regression coefficient significance; $p_{\text{E-L}}$ refers to the paired Wilcoxon early–late comparison.

Position	Muscle	MF slope	p_{slope}	$p_{\text{E-L}}$
Road	TA-R	-0.155	3.6×10^{-8}	3.1×10^{-10}
	TA-L	-0.201	4.3×10^{-12}	9.9×10^{-11}
	ST-R	-0.003	0.881	0.560
	ST-L	-0.154	6.5×10^{-11}	2.9×10^{-6}
	RF-R	-0.015	0.362	0.329
	RF-L	-0.048	0.021	0.819
	SOL-R	-0.050	0.006	6.3×10^{-5}
	SOL-L	-0.048	0.077	0.021
Frame-clench	TA-R	-0.083	4.3×10^{-6}	1.4×10^{-4}
	TA-L	0.020	0.245	0.217
	ST-R	0.023	0.174	0.694
	ST-L	-0.007	0.696	0.941
	RF-R	0.054	0.003	0.010
	RF-L	-0.127	2.0×10^{-14}	1.8×10^{-10}
	SOL-R	-0.057	0.001	0.005
	SOL-L	0.027	0.305	0.937
Time-trial	TA-R	0.021	0.243	0.064
	TA-L	-0.018	0.274	0.564
	ST-R	0.013	0.452	0.647
	ST-L	-0.066	2.7×10^{-4}	0.003
	RF-R	-0.081	8.3×10^{-8}	5.4×10^{-5}
	RF-L	-0.083	4.5×10^{-7}	1.2×10^{-4}
	SOL-R	-0.024	0.168	0.105
	SOL-L	-0.025	0.317	0.834

and ST-L), whereas the rectus femoris does not exhibit consistent statistical significance and shows limited early–late variation. Together, these results indicate that, within this configuration, fatigue-related spectral changes are predominantly associated with distal musculature involved in ankle control.

The frame-clench configuration exhibits a more heterogeneous profile. Although significant negative MF slopes are observed in TA-R, RF-L and SOL-R, other muscles do not reach statistical significance, and RF-R presents a positive slope. Early–late percentage variations are generally smaller in magnitude than in the road condition for several muscles. This coexistence of significant, non-significant and opposite-direction trends suggests a redistribution of neuromuscular demand rather than a consistent monotonic fatigue accumulation across the lower limb.

In the time-trial configuration, the most robust fatigue indicators are observed in the rectus femoris bilaterally, with significant negative slopes ($p < 10^{-6}$) and early–late reductions on the order of 4–5%. In contrast, tibialis anterior and soleus do not demonstrate statistically consistent MF trends, and their Δ MF values remain comparatively small. This pattern indicates a shift of fatigue-related spectral changes toward proximal bi-articular musculature in the more aerodynamically aggressive posture.

Overall, statistically significant MF trends are clearly position-specific and muscle-dependent. MF slopes do not show a uniform monotonic decrease across all muscles, and Δ MF values generally remain within moderate ranges. Minor discrepancies between slope direction and early–late percentage change can be attributed to non-monotonic temporal behaviour within the trial rather than contradictory fatigue signatures. RMS trends further suggest that frequency-domain changes are often accompanied by posture-specific modulation of activation amplitude, consistent with compensatory recruitment strategies rather than a global increase in neuromuscular fatigue.

Taken together, the combined interpretation of slope significance and early–late effect size supports the conclusion that posture modifications primarily redistribute fatigue-related spectral changes across muscle groups within the analysed subject, without evidence of a uniform increase in overall neuromuscular fatigue.

6.4.2. Bilateral Fatigue and Symmetry

Bilateral fatigue behaviour was further assessed by comparing MF slopes between the right and left sides. Table 6.5 reports right–left differences in MF slope for each muscle and riding position.

Table 6.5: Right-left differences in MF slope (right minus left) for the 5-minute steady-state trial. Values close to zero indicate symmetric fatigue development.

Position	TA	ST	RF	SOL
Road	+0.0198	+0.1632	+0.0560	+0.0071
Frame-clench	−0.1046	+0.0217	+0.1827	−0.0526
Time-trial	+0.0380	+0.0755	+0.0008	−0.0001

Across riding positions, bilateral differences remain generally limited, indicating a high degree of symmetry in fatigue development. Moderate asymmetries are observed for the rectus femoris and semitendinosus in selected conditions; however, these do not follow a consistent posture-dependent pattern. They are therefore interpreted as muscle-specific or subject-specific effects rather than systematic asymmetries induced by riding posture.

6.4.3. Position-Specific Observations

A more detailed inspection of the fatigue indicators reported in Table 6.3 highlights position-dependent differences that are predominantly muscle-specific rather than global.

In the road position, frequency-domain fatigue indicators are consistently negative for the tibialis anterior and soleus on both sides, indicating a clearer fatigue development in distal muscles responsible for ankle control. In contrast, bi-articular muscles show smaller and more variable frequency trends, suggesting a less uniform fatigue response across the lower limb.

In the frame-clench position, fatigue indicators appear more heterogeneous. Several muscles exhibit near-zero or slightly positive frequency slopes, particularly for the tibialis anterior and semitendinosus on one side. This behaviour suggests a redistribution of neuromuscular demand that may attenuate sustained fatigue accumulation in specific muscles, although this effect is not systematic across all muscles or sides and should therefore be interpreted with caution.

In the time-trial position, clearer fatigue signatures are observed for the rectus femoris, which exhibits consistently negative MF and MNF slopes bilaterally, indicating a more pronounced fatigue development in this bi-articular muscle. Conversely, distal muscles such as the soleus and tibialis anterior show smaller-magnitude frequency shifts compared to the road position, suggesting a partial shift of neuromuscular demand away from the ankle joint.

Overall, no riding position emerges as uniformly superior in terms of neuromuscular fatigue mitigation. Instead, changes in riding posture primarily redistribute fatigue development

across different muscle groups.

These position-dependent fatigue patterns appear coherent with the kinematic differences observed across configurations. In particular, variations in joint range of motion (ROM) may influence the mechanical work performed by specific muscle groups and the extent of cyclic muscle length changes. Positions characterised by larger hip or knee excursions are likely to increase the functional involvement of proximal bi-articular muscles, whereas configurations with greater ankle ROM may amplify the demand on distal muscles.

Consistently, the road position—associated with greater distal joint excursion—shows clearer fatigue involvement of ankle-related muscles, while the time-trial configuration, characterised by altered hip and knee kinematics, places relatively higher demands on proximal bi-articular muscles, particularly the rectus femoris. The frame-clench position exhibits intermediate behaviour, with less consistent fatigue signatures across muscles.

These findings suggest that aerodynamically optimised positions do not induce a global increase or reduction in fatigue per se, but rather modify joint kinematics in a way that redistributes neuromuscular demand across the limb. These findings provide the basis for the integrated biomechanical interpretation developed in the following section.

6.5. Discussion of Biomechanical Findings

This section discusses the biomechanical results presented in this chapter, integrating kinematic analysis, EMG activation patterns, musculoskeletal model comparisons, and fatigue indicators.

6.5.1. Posture and Kinematic Adaptations

The kinematic analysis revealed clear posture-dependent adaptations, primarily affecting the upper body. Transitions from the road to the frame-clench and time-trial positions resulted in reduced trunk inclination and substantial changes in shoulder and elbow joint angles. These changes reflect the mechanical requirements imposed by different hand support strategies and upper-body configurations.

In contrast, lower-limb kinematics exhibited relatively limited sensitivity to riding posture. Hip, knee, and ankle joint ranges of motion remained within physiologically plausible ranges for cycling and showed only moderate variations between configurations. In particular, knee joint ROM was largely invariant, indicating that lower-limb motion is primarily constrained by crank geometry and saddle setup rather than by upper-body posture.

Overall, these findings indicate that changes in riding position predominantly affect upper-body posture while preserving lower-limb kinematics associated with power production.

6.5.2. Neuromuscular Coordination Across Conditions

Cycle-averaged EMG activation patterns demonstrated a high degree of robustness across riding positions, power outputs, and cadences. When EMG amplitude was normalised, no systematic differences in activation timing or waveform shape were observed between experimental conditions.

This invariance suggests that the fundamental neuromuscular coordination strategy of cycling is preserved despite changes in posture and mechanical demand. The absence of posture-dependent timing shifts indicates that riders adapt to different configurations primarily through mechanical and kinematic adjustments rather than through altered coordination patterns.

Observed differences in EMG variability were generally modest and, in the low-power high-cadence condition, are likely influenced by experimental factors rather than reflecting clear or systematic neuromuscular adaptations.

6.5.3. Insights from Musculoskeletal Model Comparison

The comparison between experimentally measured EMG signals and musculoskeletal model predictions highlighted both limitations and strengths of inverse-dynamics-based muscle recruitment in cycling.

While the model was unable to accurately reproduce subject-specific activation timing for several muscles, particularly bi-articular ones, it consistently preserved antagonistic muscle behaviour at the joint level.

This result indicates that, despite discrepancies at the muscle-activation level, the model produces mechanically consistent joint torques. Accordingly, the musculoskeletal model is considered suitable for kinematic reconstruction and joint-level inverse dynamics, but not for quantitative prediction of muscle activation or fatigue.

6.5.4. Fatigue Distribution and Muscle-Specific Effects

Fatigue analysis revealed clear neuromuscular fatigue development, as indicated by predominantly negative MF and MNF slopes across muscles and riding positions. However, fatigue magnitude and consistency were strongly muscle-specific.

No riding position consistently minimised fatigue across all analysed muscles. Instead, posture-dependent effects manifested as a redistribution of fatigue among muscle groups. Distal muscles involved in ankle control tended to exhibit clearer fatigue signatures in the road position, whereas more proximal bi-articular muscles, particularly the rectus femoris, showed more pronounced fatigue indicators in the time-trial configuration. The frame-clench position displayed intermediate behaviour, with less consistent fatigue trends across muscles.

These findings suggest that changes in riding posture modify how neuromuscular demand is distributed across the lower limb, rather than uniformly increasing or decreasing overall fatigue. The biomechanical results presented in this chapter provide a detailed characterisation of posture-dependent kinematics, neuromuscular coordination, and fatigue development in tandem cycling. By isolating biomechanical effects, this chapter establishes a necessary foundation for the integrated comparison with aerodynamic performance presented in Chapter 7.

7 | Conclusions and Future developments

This chapter integrates the aerodynamic findings presented in Chapter 5 with the biomechanical results discussed in Chapter 6, providing a system-level interpretation of posture optimisation in Paralympic tandem cycling.

Rather than treating drag reduction and neuromuscular demand as independent objectives, the following discussion examines their interaction within a coupled aerodynamic–biomechanical–structural framework.

Although the present investigation focuses on a Paralympic tandem system, the interplay between aerodynamic optimisation and biomechanical sustainability is not exclusive to tandem cycling. The trade-off between drag reduction and joint mechanics, muscle coordination, and fatigue redistribution is a general characteristic of time-trial cycling, independent of the number of riders involved.

7.1. Aerodynamic Performance in Competitive Context

The aerodynamic analysis demonstrated that global drag area across the investigated tandem configurations ranged between 0.251 and 0.271 m². Compared to the baseline configuration v0 (0.265 m²), the best-performing configuration (v5, 0.251 m²) achieved a drag reduction of 5.3%.

At constant power output, cycling velocity scales approximately as

$$v \propto \left(\frac{1}{C_d A} \right)^{1/3}.$$

Reducing C_dA from 0.265 m^2 (v_0) to 0.251 m^2 (v_5) therefore produces

$$\frac{v_{v5}}{v_{v0}} = \left(\frac{0.265}{0.251} \right)^{1/3} \approx 1.018,$$

corresponding to an increase in speed of approximately 1.8% at equal power output.

To contextualise this magnitude, the Men's B tandem individual time trial at the 2025 UCI Para-cycling Road World Championships in Ronse covered 34.8 km and was won at an average speed of 52.02 km/h. Applying the measured aerodynamic improvement to this scenario yields an estimated time saving of approximately 43 seconds over 34.8 km at constant power.

Such a magnitude exceeds typical podium time gaps in elite competitions and confirms that the aerodynamic differences observed in Chapter 5 are competitively decisive.

The dominant aerodynamic mechanisms responsible for these gains include:

- improved pilot head alignment and upper-body streamlining,
- optimisation of trunk inclination,
- modification of inter-rider spacing, which alters wake interaction and produces non-additive drag effects.

7.2. Posture Optimisation and Kinematic Robustness

The aerodynamic gains identified above are primarily achieved through upper-body geometric modifications. The biomechanical analysis shows that while trunk, shoulder, and elbow joint angles vary substantially across configurations, lower-limb joint kinematics remain comparatively stable.

Hip, knee, and ankle joint ranges of motion remain within physiologically plausible cycling ranges and exhibit only moderate posture-dependent variations. In particular, hip joint kinematics demonstrate limited sensitivity to upper-body posture changes, indicating that the fundamental mechanics of power production are preserved even in aerodynamically aggressive configurations.

Within the analysed steady-state condition (300 W, 100 rpm, 5-minute trial), aerodynamic optimisation does not appear to inherently compromise pedalling feasibility.

7.3. Neuromuscular Coordination Stability

Cycle-averaged EMG activation profiles show that the temporal structure of muscle coordination remains robust across riding positions. After amplitude normalisation, no systematic shifts in activation timing or waveform shape are observed between road, frame-clench, and time-trial configurations.

These results do not support the hypothesis that aerodynamic positions fundamentally alter the coordination strategy of cyclic pedalling. Instead, perceived difficulty in aggressive configurations appears more plausibly linked to changes in muscular load distribution rather than to altered neuromuscular timing.

7.4. Fatigue Redistribution

Fatigue analysis of the sustained trial indicates posture-dependent redistribution of neuromuscular fatigue rather than a uniform increase.

No configuration consistently minimises fatigue across all analysed muscles. Instead:

- the road position shows clearer fatigue signatures in distal ankle-related muscles,
- the time-trial configuration exhibits more pronounced fatigue trends in proximal bi-articular muscles, particularly the rectus femoris,
- the frame-clench configuration presents intermediate behaviour.

Aerodynamic postures therefore shift mechanical demand within the lower limb rather than increasing global neuromuscular strain.

7.5. Position Feasibility: Structural and Stability Constraints

From an engineering perspective, tandem optimisation is constrained by structural and dynamic considerations.

Modifying inter-rider spacing is not a simple positional adjustment. Changes in longitudinal spacing alter wheelbase, load distribution, and global frame geometry, typically requiring the design of a dedicated frame. Furthermore, increased spacing may raise bending and torsional loads in the primary structural members, influencing torsional stiffness and potentially affecting pedalling efficiency and steering precision.

Adopting a time-trial configuration for the stoker, including aerobars, also requires substantial cockpit redesign. The rear cockpit is geometrically constrained by the pilot's position and synchronised pedalling dynamics, and sufficient structural stiffness must be ensured to maintain safe load transfer.

Beyond structural effects, geometric changes influence aerodynamic stability. Configurations optimised at zero yaw may exhibit different behaviour under crosswind conditions. Changes in posture and spacing modify side-force distribution and yaw moment characteristics, potentially affecting handling sensitivity.

Position feasibility must therefore be interpreted within a combined aerodynamic, biomechanical and structural stability framework.

7.6. Implications

The integrated results demonstrate that aerodynamic and biomechanical objectives in tandem cycling are tightly coupled but not inherently conflicting.

Aerodynamic optimisation primarily affects upper-body geometry and wake interaction, producing performance-relevant reductions in drag area. Lower-limb kinematics and coordination remain robust under steady-state conditions, while fatigue patterns shift across muscle groups without evidence of global amplification.

The main scientific contribution of this work lies in demonstrating that tandem time-trial optimisation must be treated as a multi-domain system problem, where aerodynamic interaction, structural constraints, neuromuscular coordination, and fatigue adaptation are tightly coupled yet not intrinsically conflicting.

By quantifying nonlinear wake interaction mechanisms within a configuration-consistent framework and linking them directly to biomechanical measurements, this study advances current understanding of tandem performance beyond purely aerodynamic or purely biomechanical analyses.

7.7. Generalisability of the Results

Some findings of the present work are specific to tandem cycling, while others extend to time-trial cycling more broadly.

More general findings applicable to time-trial cycling include:

- the robustness of lower-limb kinematics under moderate upper-body posture changes,

- the preservation of neuromuscular coordination timing despite aerodynamic optimisation,
- the posture-dependent redistribution of fatigue across muscle groups.

7.8. Design and Training Guidelines

The combined aerodynamic and biomechanical findings suggest that effective tandem position optimisation requires coordinated consideration of geometry, structural feasibility, and training specificity.

Aerodynamically aggressive configurations can be biomechanically sustainable, provided that riders are adequately adapted to the associated posture. Perceived discomfort in time-trial or frame-clench positions should therefore not be interpreted as intrinsic biomechanical penalties. Instead, the observed redistribution of fatigue toward specific muscle groups indicates that discomfort is likely related to conditioning history and neuromuscular adaptation.

From an engineering standpoint, configuration feasibility must also account for structural implications, particularly with respect to inter-rider spacing and cockpit stiffness requirements.

Moreover, aerodynamic optimisation performed at zero yaw does not automatically guarantee favourable behaviour under crosswind conditions. Changes in posture and spacing modify side-force distribution and yaw moment characteristics, potentially affecting stability and handling sensitivity.

Consequently, tandem optimisation should be understood as a system-level engineering process involving aerodynamic performance, structural behaviour, neuromuscular adaptation, and rider-specific training strategies.

7.9. Outlook for Future Research

Despite its integrated approach, the present study presents limitations that define clear directions for future work.

First, metabolic measurements were not included. Variables such as oxygen uptake, ventilation, and respiratory mechanics were not directly assessed, limiting conclusions regarding energetic efficiency. Future research should integrate metabolic and respiratory measurements to evaluate the interaction between aerodynamic posture and breathing

mechanics.

Second, systematic wind tunnel validation would strengthen the quantitative reliability of CFD predictions and enable controlled investigation of interaction effects through parametric variations in spacing and posture. In addition, open-field testing campaigns combining power-meter data, aerodynamic field testing methods, and fatigue monitoring would allow evaluation of long-duration neuromuscular responses under real racing conditions.

Third, aerodynamic modelling was conducted under steady-state assumptions. Future studies should incorporate quasi-static or time-resolved CFD across multiple crank angles, informed by motion capture data, to account for the dynamic nature of pedalling and refine aerodynamic prediction accuracy.

The musculoskeletal simulation performed within this thesis did not yield fully satisfactory quantitative agreement, particularly in the comparison between EMG measurements and musculoskeletal model predictions. While qualitative trends were observed, discrepancies highlight the limitations of inverse-dynamics modelling under highly constrained cycling postures. Further refinement of subject-specific modelling assumptions and dynamic parameter identification is required to improve predictive reliability.

This thesis demonstrates the value of integrating CFD, kinematics, EMG, and fatigue analysis within a consistent experimental framework. Such integration enables identification of configurations that balance aerodynamic efficiency with biomechanical sustainability, providing a structured foundation for evidence-based performance optimisation in elite Paralympic tandem cycling.

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