AN EXPLORATION OF THE BIOMECHANICAL CHAIN IN WALKING AND RUNNING

NEUROMECHANICAL FUNCTION FROM THE FEET UP TO THE TRUNK

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The path meanders into the mountain- always further, always onward
Dancing through the juniper, tiptoeing over the marsh,
searching over rock, hopping onto the tussock
Catch me, catch me- it says, and sets off with new speed!
And the summit plain lies with open doors-quietly stretching, in all directions -
It carries on- deeper and deeper in....

Stien bukter seg innover fjellet- alltid lengre, alltid videre
Danser frem gjennom eineren, lister på tå over myren,
famler ut over svaberg, hopper på en tue.
Fang meg, fang meg- sier den, og setter av sted med ny fart!
Og vidden ligger for åpne dører-stillferdig utstrakt til alle sider -
Det bærer dypere og dypere inn....

FROM STIEN, GABRIEL SCOTT, 1925; ENGLISH TRANSLATION: KH, 2025

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Abstract

Human gait and posture depend on coordinated interactions within the kinematic chain, where movement and force are transferred from the feet upward through the lower limbs to the trunk. As the base of support, the feet play a central role in initiating these dynamics, yet the extent to which changes in foot mechanics affect more proximal segments remains unclear. This thesis aimed to investigate how alterations in foot kinematics influence the motor behavior along the biomechanical chain-from distal to proximal segments-including the ankle, knee, hip, and trunk during functional lower limb tasks and locomotion.

Two methodological approaches were used to support this objective. Twelve healthy adults participated in these studies. Both used standardized surface electromyography (EMG), to assess neuromuscular activation. **Study I** compared maximum voluntary isometric contraction (MVIC) for EMG normalization in supine and standing positions, while **Study II** examined the influence of static foot postures (neutral, pronated, supinated) on muscle activation during a standardized step-up maneuver. The results revealed no systematic differences in terms of EMG amplitude between the supine and standing MVIC positions, and no major effect of foot posture on trunk muscle activation-even though more consistent effects were observed in the lower extremity.

Taken together, these findings support the robustness of MVIC normalization and suggest that foot configuration primarily modulates distal, rather than proximal, muscle activity.

This framework was then applied to individuals one year after a unilateral Achilles tendon rupture; a condition known to impair ankle function and disrupt gait propulsion. EMG was used to assess muscle activation in target muscles, while an optical motion capture system was employed to measure kinematic and kinetic variables.

Thirty-seven individuals with unilateral Achilles tendon rupture participated in **Study III,** which investigated the side-to-side differences in neuromechanical function during walking and running approximately one year after the tendon injury using synchronized EMG, 3D motion capture, and force plate analysis. This study revealed increased gastrocnemius activation and reduced ankle range of motion of the affected limb during walking, alongside with decreased plantarflexor moments during running. EMG and joint kinetics showed greater variability than kinematics, suggesting compensatory strategies that preserve overall gait symmetry.

A subset of 22 participants from the same cohort as in Study III, were included in **Study IV**, which extended the analysis along the biomechanical chain-from the feet to trunk-using bilateral EMG signals from eight muscles, joint power and moments, support moment, and joint range of motion. During walking, the affected limb showed reduced ankle and knee power, diminished total support moments, and decreased ankle range of motion, with increased gastrocnemius activation and lower EMG variability in the soleus, indicating more rigid and stereotyped motor control. Trunk-level adaptations were limited, and no consistent patterns of altered proximal coordination were observed. Individuals displayed phase- and task-specific neuromechanical deficits primarily in the distal limb, especially during walking**.** Running revealed fewer between-limb asymmetries, however with persisting ankle-level deficits and altered control strategies. Taken together, these findings highlight long-term, task-specific adaptations after Achilles tendon rupture, with distal changes-especially at the ankle-playing a key role in altered neuromechanical control during walking and running. In contrast, proximal adaptations were modest, inconsistent, and varied considerably between individuals.

The present thesis demonstrates that alterations in foot mechanics-particularly in the ankle joint- have a significant impact on distal neuromuscular control during gait, especially after an Achilles tendon rupture. In contrast, proximal adaptations were limited and variable. These findings emphasize the importance of rehabilitation strategies focused on distal tendon function and neuromuscular control, while acknowledging individual differences in motor variability. In addition, the methodological work confirmed that MVIC normalization yields consistent results across testing positions, reinforcing its suitability for reliable EMG-based assessments in both research and clinical contexts.

Keywords:
Biomechanical chain, human gait, foot posture, Achilles tendon rupture, electromyography, kinematics and kinetics, neuromechanical adaptation, motor variability

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Sammanfattning på svenska

Mänsklig gång och hållning är beroende av koordinerade interaktioner inom den kinematiska kedjan, där rörelse och kraft överförs från fötterna upp genom nedre extremiteterna till bålen. Som stödjepunkt spelar fötterna en central roll i att initiera denna dynamik, men i vilken utsträckning förändringar i fotens mekanik påverkar mer proximala segment är fortfarande oklart.

Syftet med denna avhandling var att undersöka hur förändringar i fotens kinematik påverkar motoriskt beteende längs den biomekaniska kedjan – från distala till proximala segment – inklusive fotled och upp till bål i samband med funktionella uppgifter för nedre extremitet och under gång och löpning.

Två metodologiska studier – Studie I och II-genomfördes för att stödja detta syfte. Tolv friska vuxna deltog i båda studierna, där standardiserad yt-EMG användes för att analysera neuromuskulär aktivering.

Studie I jämförde metoder för maximal viljemässig isometrisk kontraktion (MVIC) för EMG-normalisering i liggande och stående positioner, medan **Studie II** undersökte hur statiska fotpositioner (neutral, pronerad, supinerad) påverkade muskelaktivitet under ett standardiserat uppstegstest. Resultaten visade inga systematiska skillnader i EMG-amplitud mellan liggande och stående MVIC-positioner, samt inga större effekter av fotposition på bålmuskulaturens aktivering-även om tydligare effekter observerades i nedre extremiteten. Sammantaget stödjer dessa fynd robustheten i MVIC-normalisering och antyder att fotposition främst påverkar distal snarare än proximal muskelaktivitet.

Detta ramverk tillämpades därefter på individer ett år efter en unilateral akillesseneruptur-ett tillstånd som påverkar fotledsfunktion och stör gångens framåtdrivning. Elektromyografi användes för att analysera muskelaktivitet i utvalda muskler, medan ett optiskt rörelsesystem användes för att mäta kinematiska och kinetiska variabler.

Studie III inkluderade 37 personer med unilateral akillesseneruptur och undersökte sidoskillnader i neuromekanisk funktion under gång och löpning ett år efter skadan, med hjälp av synkroniserad EMG, 3D-rörelseanalys och kraftplattor. Studien visade ökad gastrocnemiusaktivitet och reducerad fotledsrörelse i den drabbade sidan under gång, samt minskade plantarflexionsmoment under löpning. EMG och ledkinetik visade större variation än kinematiken, vilket antyder kompenserande strategier för att bibehålla övergripande gångsymmetri.

Ett delurval på 22 deltagare från samma kohort inkluderades i **Studie IV**, som utökade analysen längs den biomekaniska kedjan – från fot till bål – med bilateral EMG från åtta muskler, ledkraft och moment, stödjande moment och ledrörelser. Under gång visade den drabbade sidan minskad kraft i fotled och knä, reducerade stödmoment samt begränsad rörlighet i fotleden, tillsammans med ökad gastrocnemiusaktivitet och lägre EMG-variabilitet i soleus, vilket tyder på ett mer stelt och stereotypt motoriskt mönster. Anpassningar på bålnivå var begränsade och inga konsekventa mönster för förändrad proximal koordination observerades. Individer uppvisade fas-och uppgiftsspecifika neuromekaniska underskott, främst i den distala extremiteten, särskilt under gång. Färre asymmetrier observerades under löpning, även om kvarstående begränsningar i fotledsfunktion och förändrade kontrollstrategier bestod. Sammantaget belyser dessa fynd långvariga, uppgiftsspecifika anpassningar efter akillesseneruptur, där förändringar i distala segment-särskilt i fotleden – spelar en central roll avseende förändrad neuromuskulär kontroll under gång och löpning. Däremot var proximala anpassningar måttliga, inkonsekventa och visade stor individuell variation.

Denna avhandling visar att förändringar i fotens mekanik – främst i fotleden – har en betydande påverkan på distal neuromuskulär kontroll under lokomotion, särskilt efter en akillesseneruptur. I kontrast var de proximala anpassningarna begränsade och varierade mellan individer. Fynden understryker vikten av rehabiliteringsstrategier, som fokuserar på distala senors funktion och neuromuskulär kontroll, samtidigt som individuella skillnader i motorisk variabilitet beaktas. Dessutom bekräftade det metodologiska arbetet att MVIC-normalisering ger konsekventa resultat oberoende av testposition, vilket stärker dess lämplighet för tillförlitliga EMG-baserade bedömningar både i forskning och kliniska sammanhang.

Sammendrag på norsk

Menneskelig gange og kroppsholdning er avhengig av et koordinerte samspill i den kinematiske kjeden, hvor bevegelse og kraft overføres fra føttene via underekstremitetene til overkroppen. Føttene fungerer som kroppens støtteflate og spiller en nøkkelrolle i å initiere denne dynamikken. Det er fortsatt uklart i hvilken grad endringer i fotens mekaniske egenskaper påvirker mer proksimale segmenter . Formålet med denne avhandlingen var å undersøke hvordan variasjoner i fotens kinematikk påvirker motorisk respons langs den biomekaniske kjeden – fra distale til proksimale segmenter – inkludert ankel, kne, hofte og truncus. Dette ble studert under funksjonelle oppgaver for underekstremitetene samt under gange og løping.

For å belyse dette ble to metodologiske studier gjennomført med tolv friske voksne. begge studiene ble standardisert overflate-elektromyografi (EMG) benyttet for å analysere nevromuskulær aktivering i relevante muskelgrupper.**Studie I** sammenlignet metoder for maksimal viljestyrt isometrisk kontraksjon (MVIC) for EMG-normalisering i liggende og stående stilling, mens **Studie II** undersøkte hvordan statiske fotposisjoner (nøytral, pronert, supinert) påvirket muskelaktivitet under en standardisert steptest. Resultatene viste ingen systematisk forskjell i EMG-amplitude i liggende og stående MVIC-stilling. Fotstilling hadde ingen systematisk effekt på aktivering av overkroppens muskulatur. Det ble observert noe effekt i muskulatur underekstremitetene. Samlet støtter disse funnene robustheten i MVIC-normalisering og antyder at fotstilling først og fremst påvirker distal fremfor proksimal muskelaktivitet.

Rammeverket ble videre anvendt på en gruppe personer ett år etter en unilateral akillesseneruptur – en skade som ofte fører til redusert ankelfunksjon og endret gangmønster. Elektromyografi ble brukt for å analysere muskelaktivitet i utvalgte muskler, mens et optisk bevegelsessystem ble benyttet for å måle kinematiske og kinetiske variabler.

Studie III inkluderte 37 personer med unilateral akillesseneruptur og undersøkte sideforskjeller i nevromekanisk funksjon under gange og løping, omtrent ett år etter skade. Det ble anvendt synkronisert EMG, 3D-bevegelsesanalyse og kraftplater. Studien viste økt aktivitet i gastrocnemius og redusert ankelbevegelighet på den affiserte siden under gange. Under løping ble det observert redusert plantar fleksjonsmoment. Variasjonen i EMG og leddkinetikk var større enn i kinematikken, noe som antyder bruk av kompenserende strategier for å opprettholde gangsymmetri.

Et utvalg på 22 deltakere fra samme kohort ble inkludert i **Studie IV,** og utvidet analysen til å omfatte hele den biomekaniske kjeden – fra fot til truncus. Ved hjelp av bilateral EMG fra åtte muskler samt målinger av leddkraft og leddmoment, støttemoment og leddutslag, ble det identifisert redusert kraftutvikling i ankel og kne, reduserte støttemoment og begrenset leddutslag i ankelen på den affiserte siden under gange. Dette ble ledsaget av økt gastrocnemiusaktivitet og lavere EMG-variabilitet i soleus, noe som indikerer et mer stivt og stereotypisk motorisk mønster. På truncusnivå ble det observert begrensede tilpasninger, uten konsistente mønstre for endret proksimal koordinasjon. Deltakerne viste fase- og oppgavespesifikke nevromekaniske underskudd, hovedsakelig i den distale ekstremiteten, særlig under gange. Under løping ble det observert færre asymmetrier, selv om underskudd i ankelleddet og endrede kontrollstrategier fortsatt var til stede. Samlet viser funnene langvarige og oppgavespesifikke tilpasninger etter akillesseneruptur, der endringer i de distale segmentene – særlig i ankelen – spiller en sentral rolle i endret nevromuskulær kontroll. Proksimale tilpasninger fremstod som moderate og uensartede, med markerte individuelle variasjon.

Denne avhandling viser at endringer i fotens mekanikk – særlig i ankelen – har betydelig innvirkning på distal nevromuskulær kontroll under gange, spesielt etter en akillesseneruptur. I motsetning til dette var proksimale tilpasninger begrensede og varierte mellom individene. Funnene understreker viktigheten av rehabiliteringsstrategier som fokuserer på distale seners funksjon og nevromuskulær kontroll, samtidig som individuelle forskjeller i motorisk variabilitet tas i betraktning.

Abbreviations

ABD	abduction
ADD	adduction
ATR	Achilles tendon rupture
ATRS	Achilles Tendon Total Rupture Scores
BCG	body center of gravity
CAV	coordination variablity
COM	center of mass
DF	dorsal flexion
EMG	electromyography
EV	eversion
EXT	extension
FLEX	flexion
IC	initial contact
IN	inversion
MOCAP	motion capture
MS	midstance
MUAP	motor unit action potential
OA	osteoarthritis
PF	plantar flexion
RMS	root mean square
ROM	range of motion
RSA	radiostereometric analysis
TO	toe-off
TROM	total range of motion

Definitions in short

Chain theory	A mechanically connected chain of segments-from the foot to the head-that functions as an integrated, coordinated unit
Electromyography	The study of muscle function through the investigation of the electrical signals generated within the muscles
Gravity	Gravity refers to the attractive force that pulls objects toward the center of the planet. Newton's Law of Universal Gravitation
Ground reaction force	The ground reaction force is the force exerted by the ground on a body in contact with it. This force is equal in magnitude and opposite in direction to the force that the body exerts on the ground
Kinematics	The study of motion without considering the forces that cause it
Kinetics	The study of the forces that cause or influence motion
Motor variability	The variance of movements generated by an individual under a given task condition
Muscle function	The ability of muscles to produce force, generate movement, and stabilize joints to support posture and physical activity
Neuromechanical	The interactions between the nervous system (neu-), and the musculoskeletal system (mechanical) to produce movement and maintain posture
Newton's 2nd Law	Describes the relationship between force, mass, and acceleration, stating that the force acting on an object is equal to its mass multiplied by its acceleration, (F = m*a)
Newton's 3rd Law	For every action, there is an equal and opposite reaction; when one body exerts a force on another, the second body exerts an equal force in the opposite direction
Postural control	Study of intrinsic mechanisms of the human body that counteract gravity, with a special focus on the function of the muscular system for the maintenance of balance when posture is exposed to perturbation

List of papers

This thesis is based on the following studies, referred to in the text by their Roman numerals.

I. **Huseth K, Aagaard P, Gutke A, Karlsson J, Tranberg R.**

Assessment of neuromuscular activity during maximal isometric contraction in supine vs standing body positon.

J Electromyogr Kinesiol. 2020, 50:102365.
https://doi.org/10.1016/j.jelekin.2019.102365

II. **Huseth K, Aagaard P, Gutke A, Karlsson J, Zügner R, Tranberg R.**

The effect of foot pronation and supination during vertical step maneuvers on muscle activation in selected trunk and lower extremity muscles

Submitted

III. **Huseth K, Hardarson GR, Aagaard P, Gutke A, Zügner R, Karlsson J, Nilsson Helander K, Larsson L, Brorsson A, Tranberg R.**

Interlimb differences in gait kinematics, kinetics, and muscle activation during walking and running one year after acute unilateral Achilles tendon rupture.

J Ortop Surg Res. 2025, 20 (1): 819.
doi: 10.1186/s13018-025-06221-0

IV. **Huseth K, Hardarson GR, Aagaard P, Gutke A, Zügner R, Karlsson J, Nilsson Helander K, Larsson L, Brorsson A, Tranberg R.**

Side-to-side differences in neuromechanical function inlower limbs and trunk while walking and running one-ye ar after acute unilateral Achilles tendon rupture.

Manuscript

1. Introduction

"

**It may be that when we no longer know what to do,
we have come to our real work.
And when we no longer know which way to go,
we have begun our real journey.**

WENDEL BERRY

MOVEMENT AND PHYSICAL THERAPY

Most of us rarely consider the intricate coordination required between the feet, knees, hips, and trunk that underlies human gait. We effortlessly place one foot in front of the other to walk to the store or sprint to catch a bus, without consciously considering how these segments interact to keep us upright and moving smoothly. This seamless integration of body segments is often taken for granted, at least until something disrupts it. An injury, such as a torn ligament or ruptured tendon, or the onset of a musculoskeletal condition, such as osteoarthritis (OA) can suddenly expose just how complex and finely tuned this system truly is. Only when the movement is disrupted whether by injury, illness, or degeneration-do we become acutely aware of the body's profound reliance on neuromechanical harmony to carry out even the most routine tasks.

Human locomotion is a complex process involving the integration of neuromuscular control, muscular strength and endurance, coordinated joint movements, postural balance, and biomechanical principles grounded in physics and chemistry [1, 2]. It is the product of evolutionary refinement, driven by necessity, curiosity, and adaptation [3]. What we typically experience as effortless movement is, in reality, the outcome of a finely tuned interplay between body structures and function [4].

Our body consists of 206 skeletal bones and more than 650 muscles, integrated to form chains of movement that enable activities of daily life, recreation, work and play [5]. Humans are able to adapt and coordinate their anatomy to crawl, walk, run, climb, jump, dance – each movement is an reflection of the intricate mechanics of the biomechanical locomotor chain [6].

Human movement science is fundamental to physical therapy practice and to overall health and well-being as well. Physical therapy uses movement both to assess dysfunction and as a primary intervention to restore function, relieve pain, and enhance quality of life. Through targeted exercises, manual techniques, and education, physical therapists help individuals regain mobility, prevent injury, and improve physical performance [7, 8]. Physical therapy therefore relies on the integrated and coordinated function of the body across multiple levels [9].

Analyzing human movement presents significant challenges due to its inherent complexity and the dynamic interactions between bodily systems.

One challenge is the research field of biomechanics, which can be defined as "the interdisciplinary research that describes, analyzes, and assesses human movement" [1, 10]. Also in terms of biomechanics the breakdown of study areas is multifaceted and diverse. One such area of interest is the study of locomotion [1].

MOVEMENT AND MOVEMENT CONTROL

Locomotion like any other human movement pattern is dependent upon a control mechanism with an anchorage in the body's postural control [4]. Posture can be defined as the study of intrinsic mechanisms of the human body that counteract gravity, with a special focus on the function of the muscular system for the maintenance of balance when posture is exposed to perturbation [10]. Taken together, this means that afferent input (sensory signals) is transmitted to the central nervous system, where it is processed, leading to efferent output (motor signals) that elicit the postural responses in the muscular system.

Perturbations to postural control and human gait can be described as encountering a variety of different and unexpected obstacles [11, 12]. Postural control demands are task-specific and vary accordingly to the interpretation of the nerve feedback system [11]. As such, walking and running require sufficient inherent balance control to ensure that erect posture is sustained during the propulsive motor actions. Postural control relies on previous movement experience, from early mobility to present activity [13].

Postural control also depends on what can be called *feed-forward mechanisms* that use anticipatory signals to predict how impending movements might disrupt the balance system [4]. Each individual carries a unique movement history, a learned rhythm that shapes and creates an imprint in the individual neuro-muscular system [13].

MOVEMENT AND MOTOR VARIABILITY

An individual's ability to adapt the body movements is a significant advantage and asset in everyday skill-solving and, of course, as well as recreational and professional sports [14]. Dynamic changes are supported by the plasticity of human biological systems, enabling the adaptation of each individual's actions [15, 16]. Motor variability can be defined as the variance of movements generated by an individual under a given task condition [17-19]. The variability observed across repeated movements is not merely a random and unwanted noise, but may reflect an underlying structure that takes advantage of the motor system's built-in redundancy. It is the body's ability to achieve the same task through multiple combinations of muscles and joint movements [20, 21]. This suggests that the co-existence of determinism and variability in co-ordinated behavior may have important functional relevance, rather than representing purely random fluctuations [22, 23].

An overview of sports biomechanics research has highlighted variability in movement-specifically in throwing skills, basketball shots, and locomotion-suggesting that movement and coordination variability can be functional [24]. Such functionality may enable better adaptation to environmental changes, reduce injury risk, and support adjustments in individual coordination patterns. Bartlett et al. argued that movement and co-ordination variability are, or could at least be, functional [24]. This functionality may allow for better adaptation to environmental changes, reduce the risk of injury, and facilitate alterations in individual co-ordination patterns.

Differences in walking speed and footwear, for example, have been shown to influence kinematic and kinetic patterns, with some individuals adopting to a more flexed knee joint angles and higher extensor moments at the knee, while others walk with a more extended knee joint angles and lowerknee extensor moments [(25)].

During running, limb-specific coordination variability (CAV) appears to be more relevant to injury risk than asymmetry between limbs [(22)]. Recreational runners with overuse injuries exhibited lower CAV asymmetry than uninjured controls, particularly in knee–ankle coupling during mid-stance and in most segment couplings during late stance, except for the pelvis and thigh [(22)].

Increased muscle activation variability has also been observed at the onset of trunk muscle fatigue, in both healthy individuals and those with chronic low back pain, although the latter show less variability-likely due to avoiding painful movements [(26)]. Similarly, prolonged muscle contractions that induce fatigue tend to increase motor variability [(27, 28)] In individuals with a history of Achilles tendon rupture, greater variability in foot eversion–shank rotation coupling has been documented during 47–50% of stance-coinciding with peak ground reaction force-indicating altered segmental control and potentially abnormal loading patterns [(29)].

Understanding motor variability is important within the framework of the biomechanical chain mechanism, as it influences how different body segments co-ordinate in order to maintain an erect posture and superimposed movement.

MOVEMENT AND THE BIOMECHANICAL CHAIN MECHANISM

The erect posture is part of a biomechanical chain, as described by the *chain theory of body linkage*, which views the body as a system of mechanically connected segments-from the foot to head-and this chain functions together as a co-ordinated unit [30]. The human biomechanical chain from the metatarsal joints of the feet to the most proximal cervical joint consists of more than 40 degrees of joint freedom movement. The sizeable number of joint freedoms enables the foot to adjust to various challenging surfaces, while at the same time allowing the biomechanical chain to function as a whole to maintain the body's center of gravity (BCG) within the small base of support dictated by the feet. This is achieved through muscle activation across the interconnected skeletal muscles [31].

The muscles of the biomechanical chain can be grouped by distal-to-proximal segments; those of the foot and ankle, lower leg, knee and thigh, hip joint, pelvis, and trunk respectively. The respective muscles receive neural innervation from the lumbar and sacral plexuses (L1–S4) as well as from the dorsal rami of spinal nerves and the intercostal nerves (Th7–Th11), [5].

A biomechanical chain mechanism functions similarly to a mechanical system composed of interconnected components designed to transmit force or motion from one segment to another. In mechanical systems, components like links, gears, and sprockets interact to transfer power, control motion, or transmit torque. In a biomechanical context, this chain is composed of bones, joints, muscles, and their associated neural innervation, which work together to generate and co-ordinate movements throughout the body.

The effectiveness of this muscular co-ordination is strongly influenced by the human body's bipedal posture, which repositions the BCG closer to the spine directly in front of the second sacral vertebra (S2), close to the hips [11]. This strategic alignment enables efficient weight transmission, with forces passing just ventral to the knee and ankle joints. The architecture of the femur, along with the alignment of the tibia, knee, and foot, ensures these structures remain closely aligned with the path of the center of gravity (COG), [32]. This configuration enhances balance, minimizes energy expenditure during locomotion, and supports smooth and efficient forward movement.

The anatomical components in the investigated biomechanical chain model include the skeletal bones; the phalanges, midtarsal and tarsal bones of the foot, tibia, fibula, femur, pelvis, and finally the spine. These bones are connected through a series of joints and ligaments, such as the joints of the foot and ankle: metatarsal-phalangeal (MTP), distal interphalangeal (DIP), proximal interphalangeal (PIP), tarsometatarsal, midtarsal, subtalar, talocrural, the knee and hip joints, sacroiliac (SI), and lumbar spine joints.

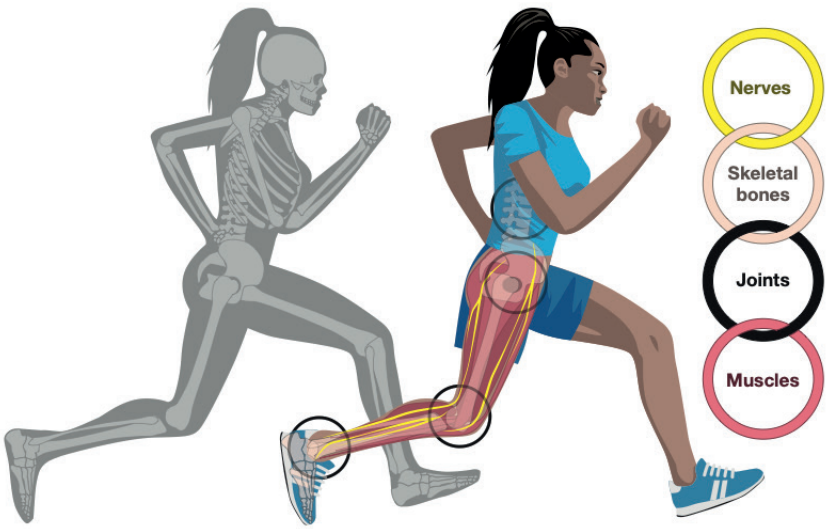

FIGURE 1. The biomechanical chain investigated in the present thesis, demonstrating the interplay between neural components, skeletal structures, joints and muscles from the foot up to the trunk.

In the biomechanical chain, forces and movements are generated and controlled by a great number of muscles, approximately 70 to 80 skeletal muscles from the intrinsic muscles of the foot to the deep stabilizers of the trunk. Together, these structures form a co-ordinated chain that enables complex motion and efficient power transmission throughout the body. For the biomechanical chain investigated in the present thesis, the involved muscles include the soleus, tibialis anterior, gastrocnemius, peroneus longus, quadriceps, sartorius, gluteus medius, rectus abdominis, external oblique, internal oblique/transversus abdominis, multifidus, and erector spinae [(5)], (Figure 1).

Understanding their role requires not only evaluating how segments are mechanically coupled, but also considering how experimental posture influences muscle activation and the corresponding electromyographic (EMG) recordings [(33, 34)]. Since neural responses are shaped by both control strategies and the mechanical context in which they occur, posture has the potential to alter muscle activation magnitude and recruitment patterns. While supine or prone positions eliminate load-bearing and postural stabilization demands, upright postures recruit trunk and lower-limb muscles in a more functionally relevant manner [(35)]. Consequently, the methodological focus of this thesis was to compare EMG normalization procedures in supine versus standing positions, as the locomotor tasks under investigation-walking and running-are inherently performed in upright stance.

BIOMECHANICS OF LOCOMOTION

A central expression of the inter-muscular walking cycle co-ordination is human gait, which can be described as the transition from a posturally unstable two-foot stance into a controlled forward fall involving progressive instability, interrupted just in time by the forward foot making contact with the ground. This results in a continuous sequence of controlled forward falls, skillfully managed in the chain events to sustain balance and forward motion [(36)].

The walking cycle consists of two primary phases-stance (60% of the cycle) and swing (40%) and these two phases alternate between the ipsilateral and contralateral lower extremities [(31)]. As the body transitions from double stance (two feet touching the ground) to single leg support (one foot touching the ground), these phases work together to propel the body forward in a coordinated and efficient manner. The kinematics of the walking cycle can be further divided into three functional phases: *weight acceptance, single limb support,* and *swing limb advancement* [32], (Figure 2). Each of these plays a critical role, with specific actions and objectives that ensure smooth, stable, and effective locomotion in an upright position [30]. The three functional phases could be further subdivided as described hereunder [30].

Weight acceptance starts with the initial contact (IC) phase, which occurs at 0-2% of the gait cycle [32]. At this point, the heel rocker mechanism is activated, and the foot makes initial contact with the ground. This phase involves impact deceleration to absorb the shock generated by the foot strike. The next phase, the *loading response*, spans from 2-12% of the gait cycle. During this time period, the contralateral limb is lifted for swing, and the body absorbs the shock from impact. The loading response also serves to provide initial limb stability and preserve forward propulsion, ensuring continued progress in movement [32].

Single Limb Support starts with the lifting of the contralateral foot for the swing phase. The *midstance* phase, which occurs between 12-31% of the gait cycle, involves progression over the stationary foot. Stability in both the limb and trunk is crucial here to maintain balance. The *terminal stance*, which occurs from 31-50% of the gait cycle, is marked by the heel rise while the contralateral foot is in contact with the ground. During this phase, the body continues to progress beyond the supporting foot, and limb and trunk stability is once again the key to successful movement.

Swing Limb Advancement includes several phases that work together to advance the limb through the cycle. The *pre-swing phase* (50-62%) is marked by toe-off and the initial contact of the contralateral limb. This phase involves positioning the limb for swing and accelerating progression. The *initial swing* that occurs between 62-75% of the gait cycle, starts when the ipsilateral foot is lifted from the floor, and the contralateral limb is in the stance phase. During this phase, the focus is on foot clearance of the floor and the advancement of the limb from its trailing position. In the *mid-swing* phase (75-87%), the ipsilateral foot is opposite the contralateral stance foot, and the swing leg's tibia becomes vertical. Limb advancement continues, and foot clearance is now maintained. Finally, *terminal swing* (87-100%) occurs when the tibia is vertical, and the foot strike is imminent. The limb has completed its advancement and is prepared to transition into the next stance phase (Figure 3).

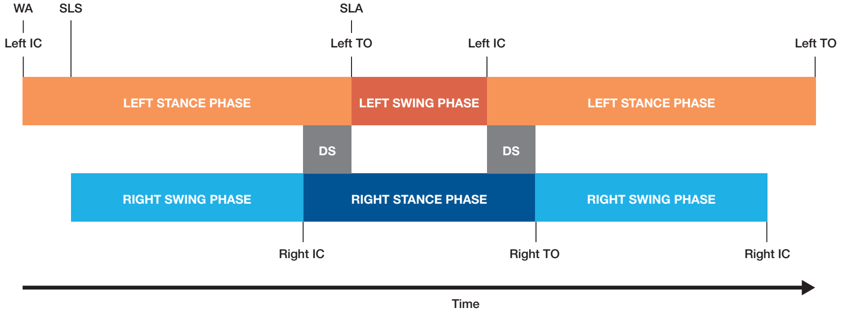

FIGURE 2. Schematic illustration of the stance and swing phases of gait, with the reciprocal action between left and right limb. WA = weight acceptance, SLS = Single Limb Support, SLA = Swing Limb Advancement, IC = initial contact, TO = toe-off, DS = double stance.

The body's subdivisions play a role in maintaining postural integrity and facilitating movement. The passenger unit (head, arms, and trunk, or HAT) comprises approximately 70% of the body weight and maintains postural integrity throughout the gait cycle. The center of mass (COM) is located approximately one-third of the distance between the hip joint center and the shoulder joint center, at the level of the 10th thoracic vertebra (Th10). Minimizing postural changes from the pelvis to the hip is essential for smooth movement. While arm swing is not strictly essential, it helps balance and aids in the overall efficiency of walking.

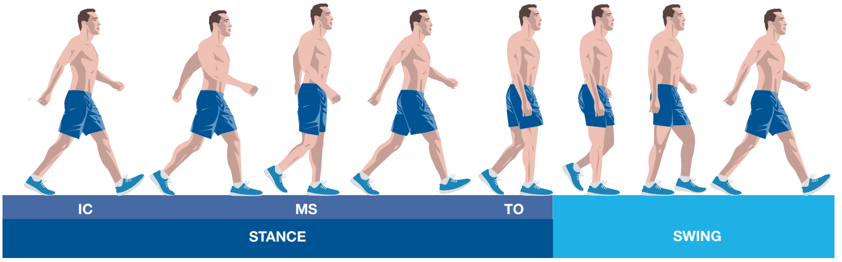

FIGURE 3. The walking cycle illustrated for the left leg. The stance phase includes initial contact (IC), mid-stance (MS), and toe-off (TO). The swing phase follows, defined as the portion of the gait cycle during which the foot is off the ground and the leg swings forward in preparation for the next step-from toe-off to the subsequent initial contact.

The locomotor unit, which includes both lower extremities and the pelvis, is critical for movement. This unit includes multiple articulations, such as the lumbosacral joint, hip joints, knees, ankles, subtalar joints, and metatarsophalangeal (MTP) joints, all working together as a smooth and well-balanced chain to propel the body forward. In terms of muscle function, the pelvis is an essential part of the locomotor unit. It advances during the swing phase and undergoes backward rotation in the terminal stance, transitioning to forward rotation in the terminal swing. This rotation facilitates efficient progression and supports the body's stability during gait.

TABLE 1. Muscles and joint/action involved in the walking cycle of the ipsilateral limb.

Gait phases	Joint	Movement	Muscles
Initial contact	hip	FLEX	iliopsoas
	knee	EXT	quadriceps
	ankle/foot	DF	tibialis anterior, extensor digitorum longus and brevis, extensor hallucis
Loading response	hip	EXT	gluteus maximus
Flat foot	knee extension	EXT	quadriceps
	ankle/foot		
Midstance	hip	EXT	gluteus maximus, hamstring
	knee work	EXT	hamstring in antagonism/synergism with quadriceps
	ankle/foot	DF	tibialis anterior
Heel-off	hip ext	EXT	gluteus maximus, hamstring
Just before end of double support	knee ext	EXT	quadriceps
	ankle/foot	PF	triceps surae , toe flex
Toe-off/pre-swing	hip	EXT	gluteus maximus, hamstring
	knee	EXT	quadriceps, hamstring
	ankle/foot	PF	triceps surae, toes flex,flexor hallux longus
Early swing	hip	FLEX	iliopsoas
Single support from other limb	knee	FLEX	hamstring
	ankle/foot	DF	tibialis anterior
Mid-swing	hip	FLEX	iliopsoas
Forward movement of swing limb	knee	EXT	quadriceps
	ankle/foot	DF	tibialis anterior, extensor digitorum longus and brevis, extensor halluces
Terminal contact	hip	FLEX	iliopsoas
Swing limb off the ground	knee	EXT	quadriceps
	ankle/foot	DF	tibialis anterior
Early advances of leading limb-preloading	hip	FLEX	iliopsoas
	knee	FLEX	hamstring
	ankle/foot	DF	tibialis anterior, extensor digitorum longus and brevis, extensor halluces

Decription of joints/joint actions and corresponding muscles during the different phases of the walking cycle. Abbreviations of movements during the walking cycle: EXT = extention, FLEX = flexion, DF = dorsiflexion, PF = plantar flexion.

Each phase and body subdivision is integral to the smooth execution of the gait cycle, highlighting the complexity and co-ordination required for efficient locomotion [36], (Table 1).

The inverted pendulum theory of walking describes how the body moves over the stance leg in a way that resembles an inverted pendulum [37]. During the single support phase of gait, the stance leg remains relatively straight, while the body's center of mass vaults over it in an arc-like motion. This "rocking" movement allows for efficient energy exchange; potential energy is converted into kinetic energy and then back again, minimizing the muscular work required for forward progression. This model helps explain the characteristic rise and fall of the COM during walking and highlights the passive, energy-conserving nature of human gait [38].

The bouncing principle of running is described by the spring-mass model, in which the body behaves like a bouncing ball or a mass supported by a spring [39]. One expression of this coordination is human gait, which can be described as the transition from a posturally unstable two-foot stance into a controlled forward fall involving progressive instability. During each step, the leg compresses upon ground contact, storing elastic energy in muscles, tendons- especially the Achilles tendon-and other connective tissues. Given its critical role in energy storage and release during the stance and push-off phases, the Achilles tendon is a key structure for efficient walking and running. For this reason, the present thesis focuses on the Achilles tendon to better understand its function and compensatory mechanisms in locomotion following injury. This stored energy in the Achilles tendon is then released during push-off, propelling the body forward. Unlike walking, where the body vaults over the rigid/straight leg (inverted pendulum), running involves a cyclical exchange between kinetic and elastic potential energies enabling efficient forward motion with minimal energy loss [40].

When walking is compared with running, the phases of movement differ significantly [41]. In walking, the stance phase makes up approximately 60% of the cycle, consisting of initial contact, midstance, and toe-off, while the swing phase makes up 40% [40]. In contrast, running features a stance phase of approximately 40%, characterized by a more pronounced push-off, and a swing phase of 60%, which includes a flight phase, where both feet are off the ground. This flight phase is a unique feature of running that does not occur during walking [31].

Ground reaction forces also vary between the two activities (Figure 4). During walking, the ground reaction force is approximately 1.2 times the body weight, with peak forces during initial contact and push-off [1]. Running, however, generates greater forces, reaching 2-3 times the body weight during foot strike. This increased force during running is also related to the higher speed and the need for more propulsion and shock absorption during the stance foot strike [39].

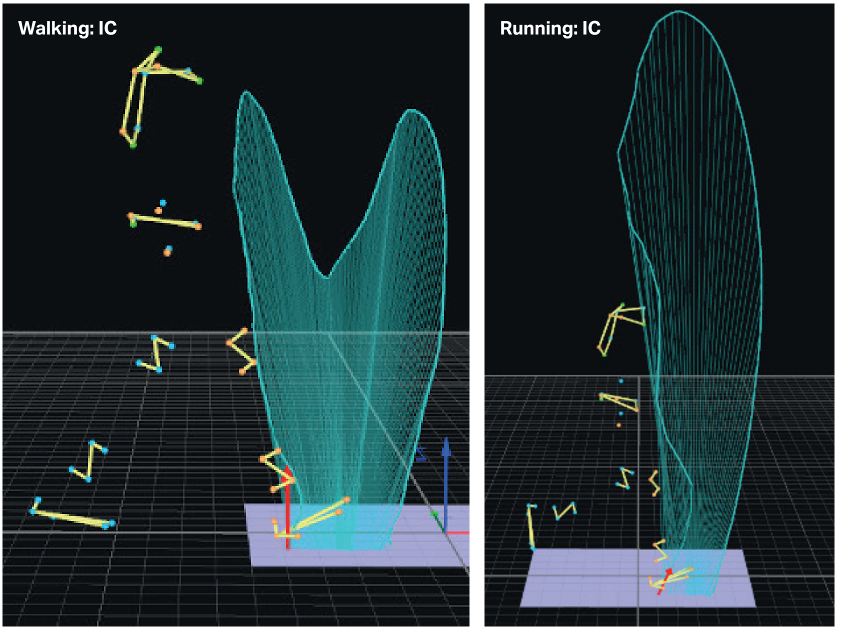

FIGURE 4. The green curves illustrate the path of the ground reaction forces (GRFs) throughout stance. The red arrow indicate the direction and magnitude of the GRF at intial contact (IC). The yellow stick figures trace the movement of the body segments, following the motion of the ankle, knee, and hip joints. Compared with walking, running produces GRFs of greater magnitude and different orientation.

Stride length and cadence further differentiate walking from running. Walking involves shorter, lower strides, while running is characterized by longer strides that are directly correlated with increased speed [42]. As for joint angles and motion, walking involves approximately 60 degrees of knee flexion during the swing phase, with a smaller range of plantar flexion (PF) and dorsiflexion (DF), [42]. Running, on the other hand, involves greater knee flexion, ranging from 90 to 120 degrees, and a larger range of PF and DF as the body needs more flexibility and mobility to generate force and propulsion [42].

Muscle activation is also more intense during running, while, during walking, muscle activation is moderate and rhythmic, with the quadriceps, hamstrings, gluteii, and shank muscles being activated to a lesser extent [42]. During running, however, muscle activation is greater than during walking, particularly in the quadriceps, hamstrings, and shank muscles, as they play key roles in absorbing impact, generating propulsion, and stabilizing the joints due to the higher forces involved. This increased muscle activity in running leads to a greater metabolic demand, making running more energy-intensive than walking. Interestingly, running is therefore metabolically more efficient than walking at higher speeds, despite its higher energy expenditure [42].

Foot strike patterns are another distinction between the two activities. Walking typically follows a heel-to-toe pattern, while running can have a more varied foot strike depending on the individual running technique. Postural control is also different; during walking, the body remains upright, with the COM staying relatively stable. During running on the other hand, there is a slight forward lean, and the COM moves more dynamically due to the flight phase.

Impact and load distribution also differ significantly between walking and running. During walking, the impact is evenly distributed, with relatively low impact on the knees and hips. Running, however, involves greater impact on the ankles, knees, and hips, which is why proper technique and muscle activation are important to minimize the injury risk. Lastly, the speed at which these activities occur is another major difference, while walking is typically performed at moderate speeds, running allows for much higher speeds, due to longer strides and faster cadence (Figure 5).

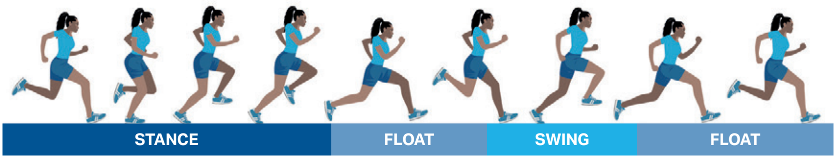

FIGURE 5. The running cycle, illustrated for the single stance phase (foot contact), flight phase (float), swing phase (forward leg movement), followed by a second flight phase before the subsequent foot strike. Unlike walking, running includes two distinct flight phases during which neither foot is in contact with the ground.

LOCOMOTION AND FEET KINEMATICS

The foot adapted to locomotion, acts as a lever adding propulsive forces to the lower limb [43].

The foot is a complex anatomical structure, composed of 26 bones and 33 joints, which together allow for a wide range of motion and adaptability. Its design enables a dynamic shift between flexibility and rigidity, largely due to the spring-like function of the foot arches [31]. These arches allow the foot to absorb shock and adapt to uneven terrain when flexible and then to provide a solid stiffer lever for propulsion during gait. This adaptability is closely linked to transitions between pronated and supinated positions. Movement within the ankle-foot complex occurs across multiple planes and axes; dorsiflexion and plantarflexion take place in the sagittal plane, rotating around a transverse axis that runs from the lateral malleolus through the body of the talus to the medial malleolus [30]. Inversion and eversion occur in the frontal plane around a longitudinal axis aligned with the length of the foot; inversion involves the plantar surface turning toward the midline, while eversion is the opposite. Adduction and abduction occur in the transverse plane, rotating around a vertical axis; abduction involves the distal aspect of a segment moving away from the midline of the foot, whereas adduction brings it closer to the midline [36], (Figure 6).

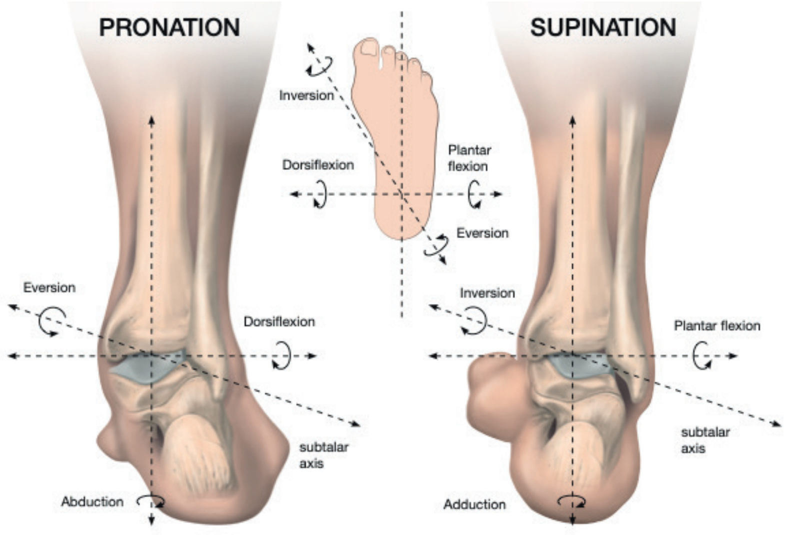

FIGURE 6. The oblique axis of ankle motion at the subtalar and midtarsal joints, along which pronation (dorsiflexion, eversion, abduction) and supination (plantarflexion, inversion, adduction) occur as coordinated triplanar movements.

Supination and pronation are fundamental foot movements that occur in response to the dynamic demands of locomotion. Supination is the combination of adduction, plantarflexion, and inversion, which results with the foot becoming more rigid, facilitating a stable push-off. In contrast, pronation involves abduction, dorsiflexion, and eversion, allowing the foot to absorb shock and adapt to the surface during weight acceptance. These movements are governed by triplane motion, with an oblique axis running in an angular direction, enabling motion across the frontal, transverse, and sagittal planes.

During the gait cycle the foot will be in a neutral to slightly supinated position during heel strike and in the phase of initial ground contact, while moving into pronation during the midstance phase, and finally moving into supination during the late terminal stance and pre-swing [44].

The range of motion for the foot and ankle complex includes 20° of dorsiflexion, 50° of plantarflexion, 20° of eversion, 10° of inversion, 15° of abduction, 15° of adduction, 35° of supination, and 20° of pronation [31, 32, 45].

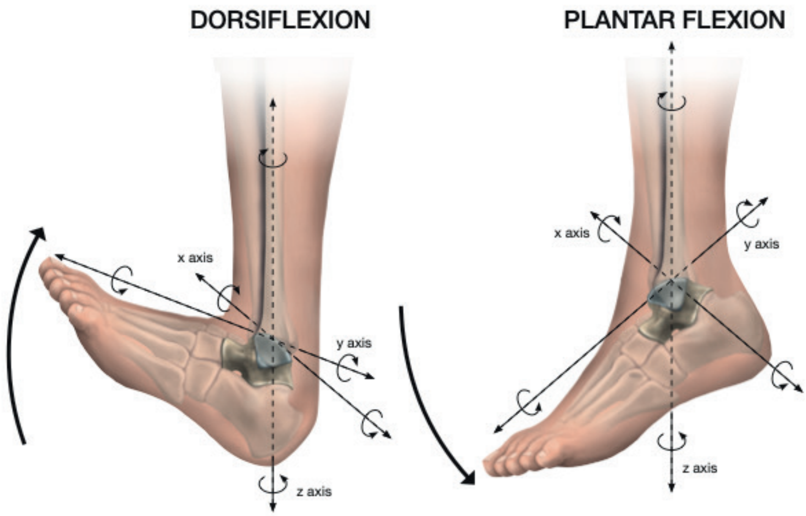

FIGURE 7. Sagittal plane motion for ankle–foot complex, depicting dorsiflexion, plantarflexion along the x-axis, frontal plane motion along the y-axis, and transverse plane motion along the z-axis.

Through the talus, the most distal bone of the lower leg, the weight from the body is transferred to the entire foot [36]. This weight is absorbed and cushioned by the pronation movement of the foot. Arguably, when the foot moves into pronation during midstance the tibia and femur rotate internally, and consequently the pelvis on the same side rotates anteriorly, causing a transient increment in lumbar lordosis [31, 46].

This coordinated sequence highlights the kinematic chain in action, where movement at one joint influences motion at other joints more proximally in the chain. In the context of locomotion, the role of the biomechanical chain is particularly evident at the ankle, where agonist and antagonist muscle pairs control dorsiflexion and plantarflexion respectively, each contributing to distinct phases of the gait cycle [31,

[47]), (Figure 7). Dorsiflexors, such as the tibialis anterior, are active during the initial contact and loading response to control foot placement, while plantar flexors, like the gastrocnemius and soleus, are key muscles during the push-off to generate forward propulsion [(32)]. When the Achilles tendon is injured, the efficiency of this system is potentially compromised [(48, 49)]. The inability to effectively plantarflex the foot will then disrupt the timing and force of push-off [(50)], which in turn will affect the mechanics of the entire kinematic chain. This could lead to compensatory movements at the knee, hip, and pelvis, and potentially exacerbate issues such as changed, i.e. either increased or decreased lumbar lordosis and placing additional strain on the lower back and postural alignment.

LOCOMOTION AND THE ACHILLES TENDON

The Achilles tendon is among the largest and strongest tendons in the human body, playing a critical role in bipedal locomotion [(51, 52)]. Evolutionarily optimized for both repetitive loading and explosive movement, it transmits forces from the triceps surae-comprising the gastrocnemius medialis, gastrocnemius lateralis, and soleus muscles-into powerful ankle plantarflexion [(53)], (Figure 8). The gastrocnemius muscle has two distinct heads-medial and lateral. The medial head, which is larger, originates from the popliteal surface of the femur just above the medial femoral condyle. The lateral head arises from the upper and posterior surface of the lateral femoral condyle and the distal part of the lateral supracondylar line of the femur. Both heads are also interviened with fibers from the subjacent areas of the knee joint capsule. They remain separate as they descend until they converge into a broad aponeurosis. The gastrocnemius is a biarticular muscle, crossing both the knee and ankle joints, and therefore contributes to movement at both levels. As it moves caudally, the gastrocnemius muscles join with the tendon of the soleus muscle on its deep surface to form the Achilles tendon. The soleus muscle, a broad and flat muscle that lies deep to the gastrocnemius muscle, originates from the posterior surface of the head and proximal shaft of the fibula, the soleal line, and the middle third of the medial border of the tibia. Unlike the gastrocnemius muscle, the soleus muscle is a monoarticular muscle, acting on the ankle joint only. Although mostly covered by the gastrocnemius, the soleus muscle becomes visible and more easily accessible just below the mid-calf on both the medial and lateral sides. Both muscles are innervated by the tibial nerve derived from spinal nerve roots S1 and S2 [45].

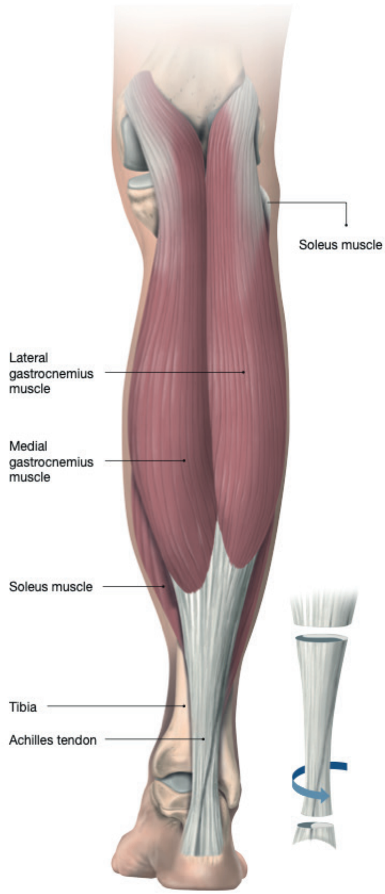

FIGURE 8. The triceps surae of the right leg, comprising the medial and lateral gastrocnemius and the soleus, arises from distinct sites: the gastrocnemius from the medial and lateral femoral condyles, and the soleus from the posterior surface of the tibia and fibula. These muscles converge into the Achilles tendon, which inserts onto the calcaneus. The right side of the illustration does also depict the rotation of the Achilles tendon.

This musculotendinous unit functions as an energy-storing spring: the tendon elongates under load during stance and recoils during push-off, which leads to enhanced propulsion and less metabolic cost of movement [2, 54, 55]. During walking and running, tendon loads range from 2.7 to 7.7 times body weight, depending on the speed and task intensity [56].

Structurally, the Achilles tendon demonstrates considerable mechanical resilience. It is composed of fascicles with a spiraling orientation and a distinctive 90° twist from the myotendinous junction to the calcaneal insertion [57-59]. This anatomical configuration, in conjunction with aponeurotic integration of the triceps surae, enables multidirectional load distribution and efficient force transfer, especially at the ankle joint level [51, 60]. The tendon's mechanical properties are tightly coupled to neuromuscular control strategies that regulate stiffness, compliance, and energy absorption during the entire gait cycle [54, 61].

Although structurally well-adapted to handle high loads, the Achilles tendon remains particularly vulnerable to injury as a result of the substantial and repetitive mechanical demands it encounters during activity, especially under eccentric loading conditions [62, 63], (Figure 9). Achilles tendon rupture (ATR) typically occurs during forceful dorsiflexion of a plantarflexed foot or during sudden acceleration or deceleration movements, such as jumping or directional changes, for instance during sports activities [62, 63]. Contributing factors often include neuromuscular fatigue, inadequate preparatory muscle activation, and age-related or subclinical degenerative changes, such as disorganized collagen structure, neovascularization, and microruptures. These changes often occur in the relatively hypovascular midportion of the tendon, a region particularly vulnerable to age-related degeneration [64, 65].

The prevalence of ATR has shown a gradual increase over recent decades across several Nordic countries. In Denmark, the incidence rose from 26.9 per 100,000 person-years in 1994 to 31.7 in 2013 [66], while in Sweden it increased from 34.3 in 2017 to 41.7 in 2021 [67]. Similarly, in Finland a rise from 17.3 to 32.3 per 100,000 person-years has been reported [68]. This trend is accompanied by an increase in the median age of patients who sustain an ATR, from 44 to 50 years between 2001 and 2012, and also with a notable rise among individuals over 60 years of age [66-68]. Recent studies also report a bimodal age distribution, with incidence peaks in middle age and in older adults [69]. Vosseller et al., reported a strong male predominance, with a male-to-female ratio of 5.4:1 [70]. Similarly, Hartman et al., in their retrospective study, found that males accounted for almost 82% of all ATR [71] and the median age of rupture has shifted upward, most probably reflecting increased participation in recreational sports among older individuals.

Irrespective of the treatment protocol (surgical or non-surgical), persistent functional deficits are commonly observed following the Acilles tendon injury [72, 73]. Tendon elongation is strongly associated with long-term reduction in plantarflexor strength, heel-rise height, and altered ankle joint kinetics [74-76]. Recent studies have further highlighted changes in tendon stiffness [77] and muscle architecture, such as shortened medial gastrocnemius fascicles [78], which, indicates remodeling responses within the triceps surae complex [71].

Compensatory neuromuscular adaptations are frequently observed after an ATR. Increased reliance on the lateral gastrocnemius and soleus is often reported, possibly reflecting reduced activation or hypotrophy of the medial gastrocnemius [79, 80]. Anatomical remodeling, including fascicle length adaptation, may, however, allow the muscle-tendon unit to function effectively under altered mechanical conditions [78].

At the joint level, ATR disrupts the coordinated distribution of mechanical work. Several studies have shown reduced ankle power and increased reliance on knee and hip extensors during walking, jogging, and dynamic tasks, such as jumping and/or landing [49, 81]. These findings could align with the concept of the biomechanical chain and the broader chain theory of body linkage, in which impairments at the foot or ankle level may influence more proximal segments in order to maintain global locomotor function.

FIGURE 9. Acute Achilles tendon rupture is typically described as a sudden blow or kick to the back of the calf, accompanied by sharp pain and sometimes a distinct snapping sound. Following the injury, walking and running are severely impaired in most cases.

Emerging evidence suggests that the post-injury adaptations extend beyond push-off mechanics and also affect early stance control [82]. Normally, the deceleration of the body's COM at initial contact is modulated by coordinated dorsiflexor, quadriceps, and hip extensor activity. Tibialis anterior controls ankle plantarflexion in an eccentric mode, while the quadriceps and gluteus absorb impact and stabilize the trunk. Post-ATR, however, altered activation pattern-including reduced support moment at initial contact, have been observed, suggesting impaired management of forward momentum [49, 79].

Human gait is not solely governed by mechanical or neuromuscular systems; psychological factors such as fear of reinjury, confidence, and emotional state can significantly influence gait and motor control, particularly during recovery after an injury [83]. These psychological influences can play a critical role in post-injury adaptation. For example, fear of reinjury is prevalent among individuals one year after ATR, with over 50% of injured patients reporting such fear; an experience that correlates with lower Achilles Tendon Total Rupture Scores (ATRS) and observable alterations in movement strategies [84].

This notion is echoed by Jónsdóttir et al., [85], who demonstrated shifts in joint power distribution during drop jumps in individuals with high fear of reinjury, despite no overt mechanical deficits. Such findings emphasize that movement alterations post-ATR are not only mechanical but may also reflect a biopsychosocial interplay between structural remodeling, neuromuscular control, and psychological readiness [86, 87]. While psychological factors are undoubtedly important in shaping movement and recovery, this thesis is specifically focused on biomechanical and neuromuscular aspects, leaving psychological influences as a valuable area for future research.

Given the Achilles tendon's central role in propulsion and load transmission during gait, its dysfunction provides a relevant context for exploring how local impairments may influence movement co-ordination throughout the kinetic chain.

Although often implicitly assumed in clinical reasoning and theoretical models, the idea that body segments interact as a part of a coordinated biomechanical chain has received limited direct investigations in the context of human locomotion. It is plausible that alterations in foot kinematics-given the foot's role as the initial point of contact with the ground-may influence neuromuscular activation and movement co-ordination in more proximal segments, such as the hip and trunk. This is particularly relevant in conditions such as ATR, where local impairments may disrupt the timing and force generation during push-off, potentially leading to compensatory strategies all along the lower limb and trunk. Such adaptations may be reflected in altered motor variability, which could serve as a mechanism to preserve functional performance and maintain postural control under changing mechanical demands such as those arising from varying physical activities, surface conditions, or movement intensities.

The present thesis was driven by an interest to examine how local changes (nonpathological and pathological) at the foot and ankle-particularly involving the Achilles tendon-influence coordination along the kinematic chain during walking and running. Despite its central role in propulsion, the broader impact of Achilles tendon dysfunction on whole-body movement remains poorly understood, in part due to the complexity of studying multi-segment interactions.

Through these studies, the overarching aim was to address these gaps and contribute to a better understanding of locomotor adaptation and recovery.

2. Knowledge gaps

"

Løpe fra meg selv? – Nei, det går ikke an.
Run away from myself? – No, that cannot be done.

PER GYNT, HENRIK IBSEN

There is limited evidence comparing EMG normalization strategies across postural conditions, and it remains unclear how the test positions supine versus standing affect MVIC values for lower extremities and trunk muscles.

The influence of foot posture, i.e. neutral, pronated, or supinated in lower extremity and trunk muscle activation during functional tasks has not been clearly established, limiting the understanding of how distal alignment affects segmental coordination.

There is insufficient knowledge of how an ATR affects lower-limb joint mechanics and muscle activation patterns during the stance phase of walking and jogging, particularly with regard to persistent between-the-limbs asymmetries after tendon rupture.

The distribution of neuromechanical adaptations along the kinematic chain from the ankle to the trunk following an ATR remains poorly understood, especially in terms of how altered distal function influences proximal joint coordination and supports strategies during walking and running.

3. Aims

"

Two roads diverged in the woods, and I–.
I took the one less travel by,
And that has made all the difference.

ROBERT FROST

The overall goal of this thesis is to advance the understanding of the biomechanical chain mechanism from the feet up to the trunk. This includes investigating how foot kinematics influence muscle activation in the lower extremities, pelvis, and trunk, alongside kinematic and kinetic measurements of interconnected body segments during various functional activities. Methodological investigations address how the position used for MVIC, standing versus supine, affects EMG normalization, thereby influencing the interpretation of muscle activation patterns in relation to foot kinematics.

STUDY I

To compare the MVIC EMG normalization method for selected lower extremity and trunk muscles when using standardized supine versus standing test positions.

STUDY II

To compare the activation of selected trunk, pelvic and lower extremity muscles during neutral, pronated, and supinated stance conditions during a standardized vertical step maneuver.

STUDY III

To investigate the limb-to-limb differences in kinematics, kinetics, and muscle activation patterns of the lower extremities during the stance phase of walking and jogging, one year after an ATR.

STUDY IV

To investigate side-to-side differences in muscle activation, range of motion, joint power, joint moment and support moment during walking and running one year after an acute ATR with specific emphasis on segmental adaptations within the ankle–knee–hip complex and their distribution along the kinematic chain up to the trunk.

4. Ethical approvals

"

Umveg er jamt, til utrygg ven,
Um midt i bygdi han bur.
Men beinvegar gjeng till den gode venen,
Um han er langt av leid.
The road is long to a false friend,
Even if he lives next door.
But the way is short to a true friend,
Though he may live far more.

HÅVAMÅL (DEN ELDRE EDDA)

All studies were approved by either the Regional Ethical Review Board in Gothenburg (Studies I and II) or the Swedish Ethical Review Authority (Studies III and IV), (Table 2). All procedures were conducted in accordance with the ethical principles outlined in the Declaration of Helsinki.

TABLE 2. Ethical approvals of Studies I-IV.

Studies	Ethical board	Date of approval	D-nr
I, II	Regional ethical review board, Gothenburg	14-09-25	514-14
III	Swedish ethical review authority	19-11-20	2019-05457
IV		22-03-18	2022-00921-02

5. Introduction to biomechanical tools

"

Not all those who wander are lost.

J.R.R. TOLKIEN

This thesis employed three primary biomechanical assessment modalities: EMG, kinematics, and kinetics. All measurements were performed at the Gait Laboratory of the Orthopedic Research Unit, Sahlgrenska Academy, University of Gothenburg.

ELECTROMYOGRAPHY (EMG)

Electromyography (EMG) was used to evaluate neuromuscular activity during movement. EMG evaluates muscle function by recording the electrical signals generated within muscle fibers in response to neural stimulation. This technique provides insights into motor unit activation patterns, timing, and coordination.

Recording of EMG can either be conducted intramuscularly, using needle electrodes, or using surface electrodes (sEMG) applied directly to the skin [10, 88]. The latter approach is most frequently used in biomechanical investigations [89, 90]. These recordings capture the electrical activation produced by muscle contractions, which originate at the level of the motor unit. A motor unit consists of a single motor neuron and the muscle fibers it innervates. When a motor neuron sends an action potential, it triggers an electrical response in the muscle, known as a motor unit action potential (MUAP). This process initiates muscle contraction, which is the basis of EMG recordings [28], (Figure 10).

SURFACE EMG (sEMG)

Surface EMG is a non-invasive method for recording electrical activity from muscles using electrodes placed on the skin. Before applying the bipolar sets of electrodes, proper skin preparation is essential to ensure accurate signal acquisition. This typically involves shaving hair at the area with a razor and cleaning the skin with an alcohol rub to remove debris [91], (Figure 11).

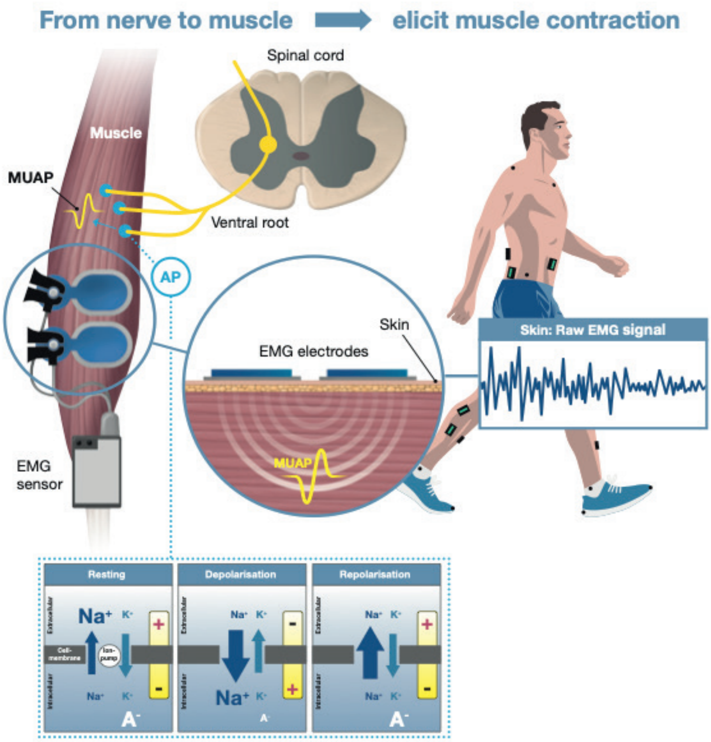

FIGURE 10. Schematic representation of the pathway from neural activation to raw surface EMG signal. The process begins with motor commands in the ventral horn of the spinal cord, traveling via the ventral root and alpha motor neuron axon to the neuromuscular junction. This triggers motor unit action potentials (MUAPs), which sum across multiple motor units to produce a complex signal in the muscle tissue. Surface electromyography (bipolar sEMG) detects this summed activity at the skin surface, producing a raw EMG signal that contains motor unit action potentials (MUAPs), along with noise and artifacts.

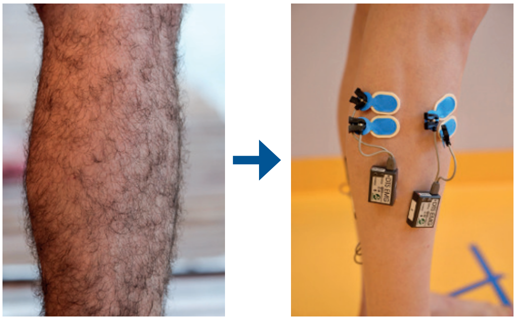

FIGURE 11. Skin preparation: Hair is first removed with a razor, after which the area is cleaned with an alcohol swab to eliminate skin debris and oils, ensuring optimal electrode–skin contact.

Electrode application follows a standardized muscle map to ensure accurate placement over the target muscle [92-94]. The procedure sampling process includes:

- Recording in rest to capture baseline muscle activity
- Signal check to ensure the quality and integrity of data
- Crosstalk check to detect interference from adjacent muscles
- Maximum voluntary isometric contraction (MVIC) recording as per established protocol
- Target muscle action recording during specific tasks, also according to a specific protocol

These procedures may need to be repeated to ensure the highest available data reliability and consistency.

sEMG SIGNAL PROCESSING

EMG signals can be affected by various sources of noise, including motion artifacts from electrode or cable movement, poor skin-electrode contact, electrical interference from surrounding equipment or lighting, and cross-talk from nearby muscles.

Signal processing of sEMG involves both analog and digital stages [95-97]. Signal processing is essential to maximize the capture and representation of muscle activation within the sampled data. Analog processing occurs at the electrode-skin interface and includes capturing the raw sEMG signal, signal amplification and applying an analog bandpass filter to reduce electrical and mechanical (movement) noise and enhance signal quality. Digital processing is performed using highly specialized software. This typically involves a digital linear envelope, which includes high-pass filter to remove low-frequency noise (movement artifacts), rectification to convert all signal values to positive, a low-pass filter, often implemented as a root mean square (RMS) moving window, to smoothen the signal (Figure 12).

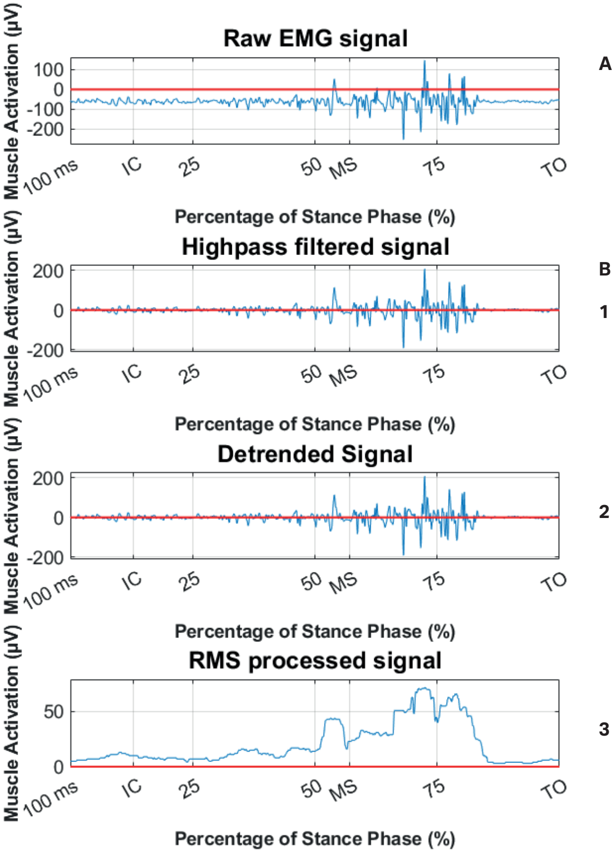

FIGURE 12. A: The analog raw EMG signal detection during the stance phase of lateral gastrocnemius muscle (one single trial) during stance phase of gait. B: the digital filtering process from 1) high pass filter 2) the detrending process to 3) the RMS process to the final signal output.

NORMALIZATION

To ensure meaningful comparisons across individuals or conditions, normalization of EMG signals is critical. This can be applied in different ways, but most used is amplitude normalization to a reference value, such as the peak EMG signal measured during a MVIC [98]. Task-specific normalization based on activity relevant to the muscle function being studied is another way of performing a signal normalization.

MOTION CAPTURE SYSTEM (MOCAP)

A three-dimensional optical camera system that uses reflective markers attached to well-defined anatomical landmarks, motion capture system (MOCAP) given body movements (kinematic action) can be recorded and analyzed (Figure 13). The markers are applied to the skin according to the given marker model, which refers to a specific algorithm [99], (Figure 14).

The MOCAP is synchronized with a force plate system that registers the external forces (kinetic actions) affecting the musculoskeletal system during the survey. This is done by performing weight-bearing movements on instrumented force plates either in the form of walking, jogging, jumping or hopping.

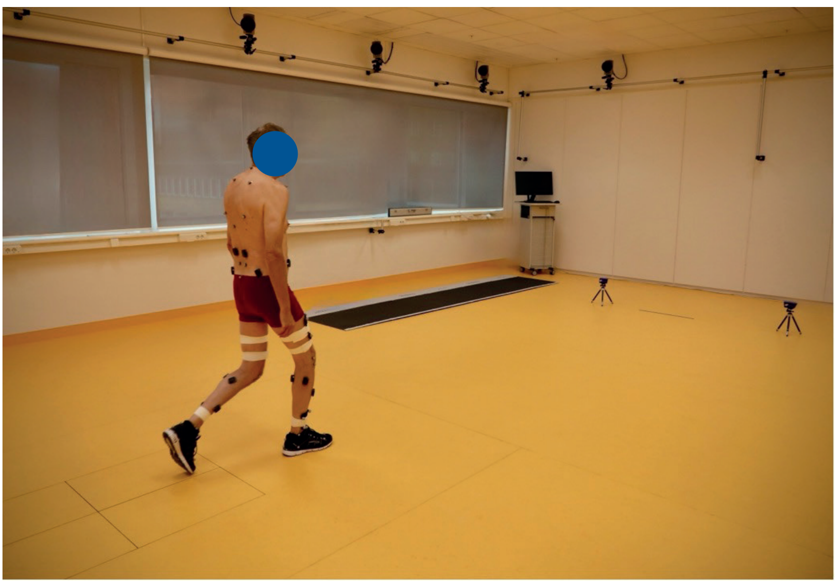

FIGURE 13. From the gait lab, the motion capture system with cameras and force plates. A participant is equipped with surface EMG sensors and reflective markers and is walking on the 10-meter stationary walkway, which is equipped with the timing system.

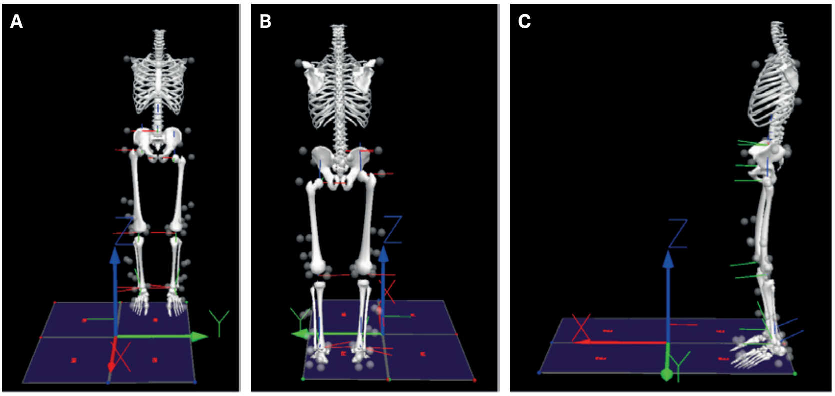

FIGURE 14. The biomechanical six degree of freedom (6-DOF) model with reflective markers on anatomical landmarks and skeletal reconstruction shown in (A) frontal, (B) dorsal, and (C) lateral views. The laboratory coordinate system (X: red, Y: green, Z: blue) defines the global reference system. Force plates embedded in the walkway capture ground reaction forces, which together with kinematic data from the marker model are used to calculate joint moment and power across the kinetic chain by inverse dynamics.

The vertical and horizontal forces exerted by the foot onto the force plate allow for quantification of the opposing ground reaction forces.

At the same time, joint angles and body segment positions are recorded by the MOCAP, which tracks reflective markers in the global Cartesian coordinate system (x, y, z). These axes correspond to the anatomical planes of motion: sagittal (forward-backward), frontal (side-to-side), and transverse (rotational), (Figure 15).

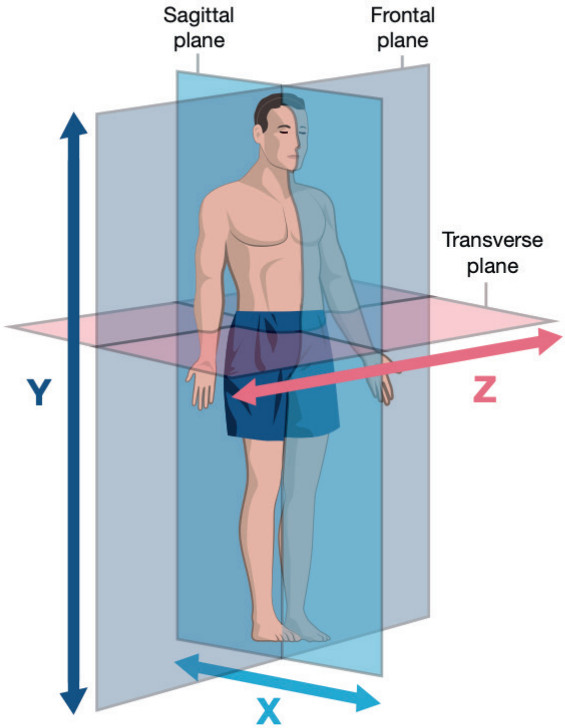

FIGURE 15. Cartesian coordinate system with the three axes (x, y, z). The sagittal plane is defined by the x–z axes (i.e., flexion and extension), the frontal plane by the y–z axes (i.e., abduction and adduction), and the transverse plane by the x–y axes (i.e., rotation). This system provides the spatial reference for describing human movement.

This combined approach-integrating kinematic data from motion capture and external force data from force plates-enables the calculation of joint kinetics through inverse dynamics calculations [1].

Kinematics refers to the study of movement without consideration of the forces that produce the motion. In the present project, segmental movement was captured using the MOCAP, which relies on high-speed cameras to track reflective markers placed on anatomical landmarks. This set-up enables precise measurement of joint angles, segment orientation, and whole-body motion yielding a detailed representation of movement patterns.

Kinetics, in contrast, focuses on the forces underlying motion. Central to the kinetic analysis is Newton´s 3rd law (Figure 16), which states that for every action, there is an equal and opposite reaction. Ground reaction forces (GRFs), measured via force plates, play a critical role in this analysis. When combined with kinematic data and known segmental properties (e.g., mass and inertia), GRFs allow for the calculation of internal joint forces and moments through inverse dynamics.

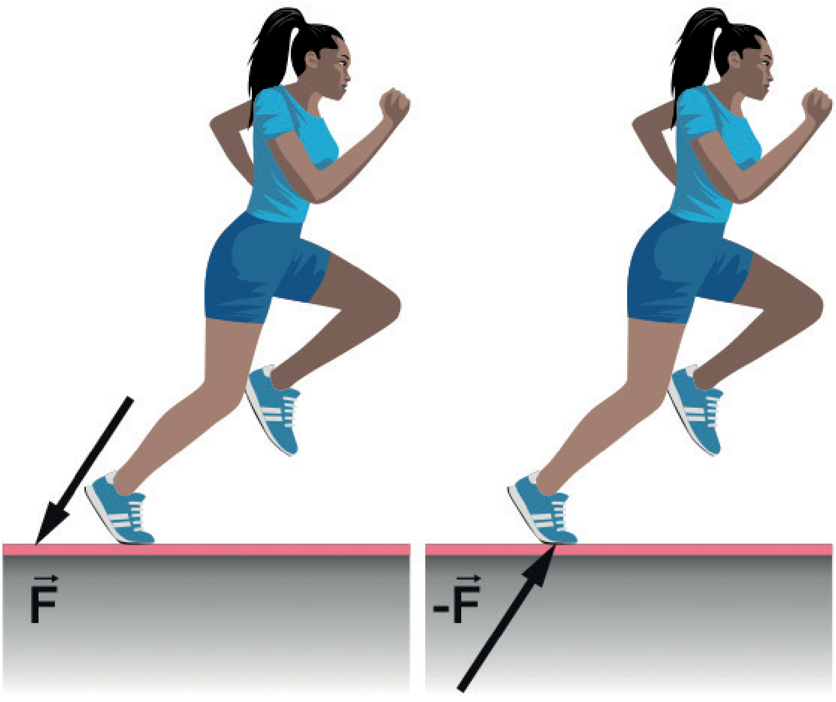

FIGURE 16. Newton's 3rd law. (a) The runner exerts a force downward and backward onto the ground. (b) In response, the ground applies an equal and opposite reaction force upward and forward, propelling the runner forward.

Inverse dynamics represents a computational method used to estimate internal joint forces and moments based on measured motion patterns (joint angles) and external forces. By combining kinematic data (e.g., joint angles and segment accelerations) with GRFs and known segmental properties (mass and inertia), inverse dynamics applies

Newton's 2nd (Figure 17) and 3rd (Figure 16) laws to work backward from movement to determine the net mechanical demands placed on each joint. This approach is essential for understanding joint loading during walking and running.

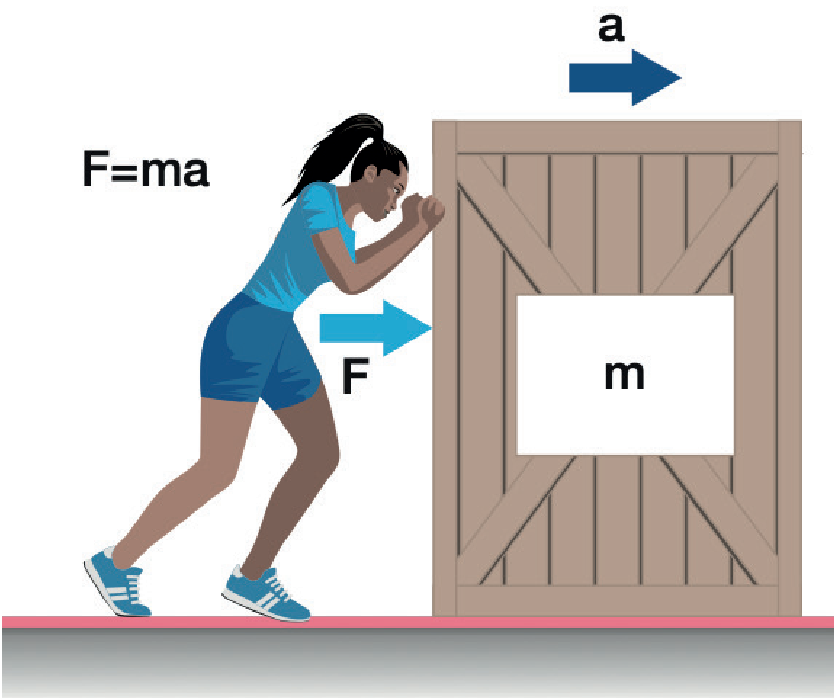

FIGURE 17. Newton's 2nd Law of Motion. A force (**F**) applied to a crate of mass (**m**) produces an acceleration (**a**) in the direction of the applied force, consistent with the relationship **F = ma**. According to Newton's 2nd law, the net force acting on an object is equal to the product of its mass and acceleration. In inverse dynamics, this principle provides the foundation for calculating net joint forces and moments: the linear acceleration of each segment's center of mass is related to the net force acting on it, while the angular acceleration is related to the net moment. By combining measured ground reaction forces with segmental kinematics, these equations yield the internal net joint moments that reflect the combined action of muscles and passive structures across the joints.

These integrated biomechanical measures-muscle activation, movement patterns, and force generation-form the foundation of the analyses performed in this thesis (Table 3).

6. Data collection; Studies I-IV

"

That's the thing about running: your greatest runs are rarely measured by racing success. They are moments in time when running allows you to see how wonderful your life is.

KARA GOUCHER

METHODS

The studies in this thesis are divided into two main categories: Studies I–II focused on methodological development, including EMG normalization and foot position during movement tasks. Studies III–IV explored the biomechanical consequences of an ATR on walking and running, with particular attention to how such injuries affect the biomechanical chain from the foot to the trunk.

TABLE 3: Biomechanical tools and variables.

	Study I	Study II	Study III	Study IV
ELECTROMYOGRAPHY				
MVIC -supine vs standing	✓			
%MVIC-mean amplitude foot kinematic		✓		
EMG%-during stance periods IC-MS, MS-TO			✓	✓
MOCAP				
ROM for stance periods: IC-MS, MS-TO			✓	
Joint positions at IC, MS, TO			✓	
ROM for IC-TO				✓
Joint moment mean during periods: IC-TO (Nm/kg)				✓
Joint moment at IC, MS, TO (Nm/kg)				✓
Joint power during IC-TO (watt/kg)				✓
PHYSICAL ASSESSMENT TOOL				
Navicular drop test		✓		
PROMs				
EQ-5D			✓	✓
PAS			✓	✓

MVIC = maximum voluntary isometric contraction, %MVIC = EMG amplitude normalized to maximum voluntary isometric contraction (100%), EMG % = detrended EMG expressed relative to %MVIC, IC = initial contact, MS = midstance, TO = toe-off, IC-MS = initial contact to midstance, MS-TO = midstance to toe-off, ROM = range of motion, PROMs = patient reported outcome measures, EQ-5D = EuroQol 5 Dimensions, PAS = physical activity scale

TABLE 4. Description of electrode placements for the EMG sampling performed in Studies I-IV.

Muscle	Electrode placement description	Studies
Tibialis anterior	Lateral to the medial shaft of tibia, approximately 1/3 of the distance between knee and ankle, over the largest muscle mass, in slightly oblique direction	I, II, III, IV
Peroneus longus	A line is drawn between the head of the fibula and the lateral malleolus, medial to muscle belly of the tibialis anterior muscle	II
Soleus medialis	Just under the belly of gastrocnemius, a line is drawn between the medial side of the Achilles tendon insertion and the head of the fibula	II, III, IV
Medial gastocnemius	Parallel to muscle fibers, just distally to the knee, 2 cm medially to the midline of the dorsal aspect of the shank	III, IV
Lateral gastrocnemius	Parallel to muscle fibers, just distal to knee and 2 cm laterally to midline of the dorsal aspect of the shank	III, IV
Vastus lateralis	Appoximately 2 cm medial from superior rim of patella, in an oblique direction (ca 55 degrees)	IV
Adductor longus	Medial aspect of thigh, approximately 4 cm from the pubic bones, in an oblique direction	I, II
Gluteus medius	Approximately 1/3 of the distance between the iliac crest and greater trochanter, anterior to gluteus maximus muscle	I, II, IV
Rectus abdominus	Approximately 3 cm superior to umbilicus and 2 cm lateral to midline	I, II, IV
External oblique	Direct in a lateral line from umbilicus. Approximately 12-15 cm, directly above the anterior superior iliac spine, halfway between the pelvic crest and the ribs, in a slightly oblique angle	I, II, IV
Internal oblique/ Transversus abdominis	Approximately 2 cm medial and inferior to anterior superior iliac spine (SIPS), on a line midway between SIPS and pubis, just superior to the inguinal ligament	I, II
Erector spinae (lumbar)	Laterally to the spinal process of L3, approximately 2 cm from the spine	II, IV
Multifidus (lumbar)	Laterally to L5 spinal process, medially to spina iliaca posterior superior (SIPS)	II

TABLE 5. EMG recorded in Studies I–IV.

EMG target muscles	Study I	Study II	Study III	Study IV
Tibialis anterior	✓	✓	✓	✓
Peroneus longus		✓		
Medial gastrocnemius			✓	✓
Lateral gastrocnemius			✓	✓
Medial soleus		✓	✓	✓
Vastus medialis				✓
Gluteus medius	✓	✓		✓
Adductor longus	✓	✓		
Rectus abdominis	✓	✓		
External oblique	✓	✓		✓
Internal oblique/Transversus abdominis	✓	✓		
Erector spinae (lumbar)		✓		✓
Multifidus (lumbar)		✓		

STUDIES I-II

In Studies I and II, EMG data were collected using a Noraxon Telemyo desktop (16 channels) and a Noraxon Telemyo belt system (8-channel) Scottsdale, USA, wireless receivers EMG system using a bandwidth filter of 10-500 Hz. Pre-amplification of all signals was performed with a baseline noise of <1μV RMS, input impedance of >100 MΩ and a common mode rejection ratio of >100 dB. Data sampling frequency was 1500 Hz using a 16-bit external A/D converter.

In both studies, surface electrodes (AMBU Blue Sensor N, Copenhagen, Denmark) were applied, and post-hoc processing of EMG signals was conducted using Noraxon MR3 software (version 3.8, Scottsdale, USA).

In Study I, two positions were compared for the performance of the MVIC EMG normalization method: supine and standing. For each target muscle, participants performed three consecutive MVICs, each lasting three seconds, with a three-second rest between repetitions. A one-minute rest period was provided between testing of different muscle groups to minimize fatigue. Throughout the testing protocol, participants were guided with standardized verbal encouragement. External static resistance was standardized and applied both manually by the examiner and with adjustable straps to ensure consistent resistance to the tested muscles (Figure 18).

A **B**

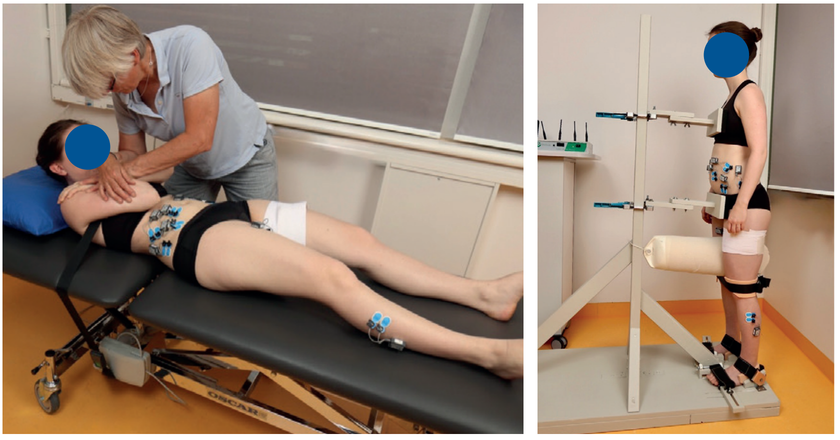

FIGURE 18. The two MVIC positions; (A) supine and (B) standing investigated in Study I.

Surface electromyographic (sEMG) activity was recorded bilaterally from six target muscles: tibialis anterior (TA), gluteus medius (GIM), adductor longus (ADDI), rectus abdominis (RA), external oblique (OE), and the internal oblique/transversus abdominis (IO/TrA) complex [91, 93, 94], (Tables 4,5). Bipolar surface electrodes with a 20 mm inter-electrode distance were placed with the participant in a standing position. Electrode placement followed established protocols as described in Table 4, [91, 93, 94].

Prior to electrode placement, the skin was prepared according to the standard procedure to reduce impedance and improve signal quality [92].

The electrodes were connected via wires to Noraxon DTR sensors, which wirelessly transmitted EMG data to a central receiver unit. Before data acquisition, a thorough signal quality check was conducted to confirm signal stability and rule out crosstalk from adjacent muscles. This was achieved through visual inspection of real-time EMG signals during light-to-moderate voluntary contractions of each target muscle.

In Study II, muscle activation of selected trunk, pelvic, and lower extremity muscles was measured during neutral, pronated, and supinated stance conditions in a standardized vertical step maneuver (Tables, 4,5). The test zone was marked with tape lines corresponding to each participant's individually determined neutral stance, ensuring a reproducible foot starting position before initiating the step ascent action (Figure 19).

FIGURE 19. The test zone is marked with tape lines for the foot positions studied in Study II. The tape lines corresponded to each participant's individually determined neutral stance, ensuring a reproducible foot starting position before initiating the step ascent action.

During the step ascent test, the stance foot was positioned in three conditions-neutral, pronated, and supinated-using individualized EVA wedges (35 Shore hardness), [100] to ensure comfort. Pronation was defined as a lateral lift, where the 5th digit was just off the ground, while supination was defined as a medial lift, where the 1st digit was just off the ground. Four trials were performed in each step condition for the right and left leg, respectively. Each participant then completed the task in a fixed sequence: (i) right neutral, (ii) left pronated, (iii) left neutral, (iv) right pronated, (v) right supinated, and (vi) left supinated. Standardized instructions were given to the participants prior to performing each step procedures. In addition, verbal cues were given on when to lift the stance foot up onto the bench and when to return the foot to the ground. A full step was defined as when the participant's contralateral foot left the ground until the same foot was placed on the bench, with a time interval of approximately 1.5 seconds.

Participants were instructed to lift one foot onto a bench and perform a single upward step, simulating a stair ascent. The bench was positioned 0.325 m in front of the starting foot, with a height of 0.445 m. Participants were instructed to maintain a forward gaze at a fixed point 3.0 m ahead throughout the movement (Figure 20).

Each participant completed the step ascent in a standardized sequence of six stance conditions. Four trials were performed for each foot position for each leg. Standardized instructions and verbal cues guided the participants during the task. A full step was defined from the moment the contralateral foot left the ground until it was placed on the bench, lasting approximately 1.5 seconds.

Bipolar, bilateral EMG signals from the target muscles were recorded (Tables 4, 5). Digital time markers were manually applied to define the relevant phases of the step action, specifically from heel lift to flat foot contact. To calculate the normalized EMG amplitude values (%MVIC), MVIC were initially performed for all muscle groups for 3–5 s using a specially designed test apparatus.

A: start B: step

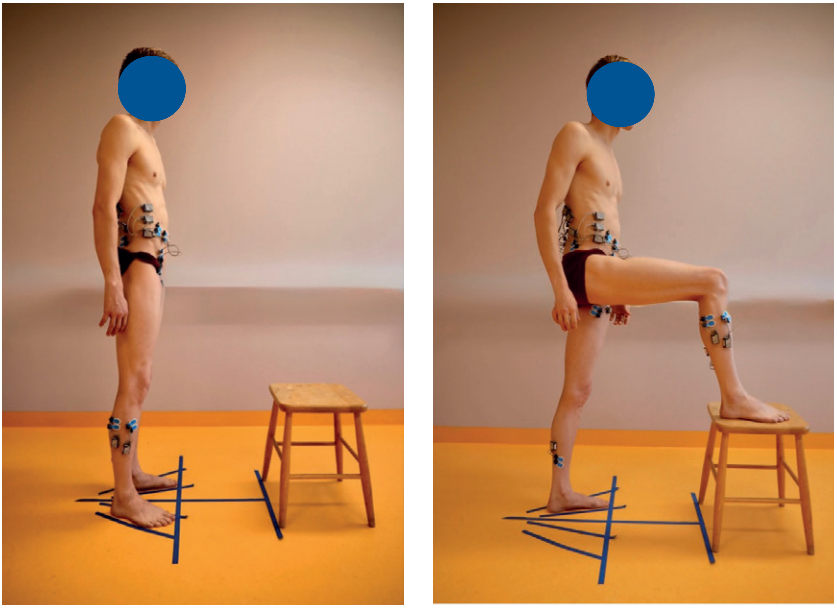

FIGURE 20. The step condition showing the left limb in stance (to be analyzed) and the right limb performing the step maneuver in Study II.

NAVICULAR DROP TEST

To classify foot posture and define pronation level, the Navicular Drop Test (NDT) was used as a clinical assessment tool in Study II.

NDT [101-103] is a static foot assessment used to evaluate the degree of pronation by measuring the vertical displacement of the navicular tuberosity from a neutral to a relaxed standing position (Figure 21). It is designed to reflect sagittal plane movement of the navicular bone, offering insight into foot posture. The test is performed with the patient standing in full weight-bearing, first positioning the foot in subtalar joint neutral. The most prominent point of the navicular tuberosity is marked and its height from the floor is measured. The patient is then asked to relax into their natural stance, and the new height is recorded. The difference between the two measurements represents the navicular drop. The test can also be conducted in reverse or by marking the positions on an index card placed along the medial side of the foot. Interpretation of results typically categorizes a drop of more than 5 mm as indicative of a supinated foot, 5–9 mm as neutral, and 10–15 mm as pronated.

A B C

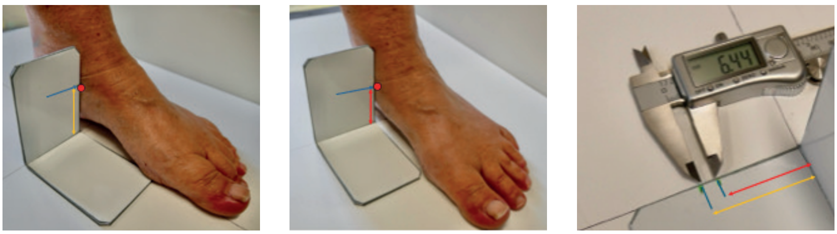

FIGURE 21. The navicular drop test. (A) Foot with the subtalar joint in neutral, with the navicular tuberosity marked (red circle). (B) Foot in relaxed stance with the navicular tuberosity marked (red circle). (C) The difference in vertical distance (mm) between the two positions of the navicular tuberosity represents the navicular drop, which can be used as an indicator of a pronated or supinated foot posture.

In Studies I and II identical off-line signal processing of the EMG signals was performed in the following order: (i) removing ECG components, (ii) high pass filtering of all raw EMG signals using a 4th order zero-lag Butterworth filter with a cut-off frequency of 2 Hz, (iii) full-wave rectification and low pass filtering using a symmetric moving RMS window (100ms time constant) and (iv) normalizing the filtered EMG amplitudes relative to peak EMG amplitude recorded during MVIC processed using identical filtering procedures.

STUDIES III-IV

In Studies III and IV, EMG data was recorded using the Delsys Trigno Research wireless system with Trigno Avanti electromyography sensors (Delsys Inc., USA). The sensors have built-in electrodes with an interelectrode spacing of 10 mm, the contact material is 99.9% silver. The system operated at a sampling frequency of 2000 Hz, recording muscle activation through a mechanical bandpass filter (25-450 Hz). The signal was reamplified with a baseline noise of 750 nV, an input impedance of >1 Gohm//20pF, and a common mode rejection ratio of >80 dB.

Kinematic and kinetic qualities were recorded using the OTS system from Qualisys AB (Sweden) with 16 infrared Oqus 7 cameras synchronized with two Miqus video cameras sampling frequency of 250 Hz, and OPTIMA High-Performance Series force plates sampling frequency of 1000 Hz from Advanced Mechanical Technologies (USA) were used, with post-hoc processing in Visual 3D (HAS-Motion, Canada). The EMG and OTS systems were synchronized.

In Study III, side-to-side differences between the affected and unaffected limbs were assessed for EMG, kinematics, and kinetics during stance phase for both walking and running. Stance phase was divided into three events: initial contact (IC), midstance (MS) and toe-off (TO) for further analysis. IC and TO were identified from the force plate data as the time points when the foot first contacted and subsequently left the plate, respectively, while MS was defined as the instant when the COG passed over the COP (Figure 22).

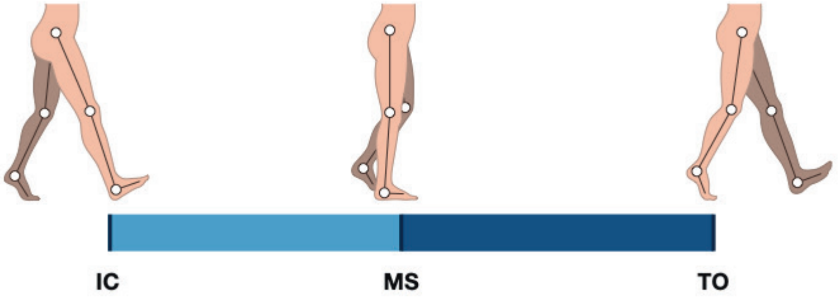

FIGURE 22. Illustration of the stance phase events as defined in Studies III and IV, including initial contact (IC), midstance (MS), and toe-off (TO).

A B

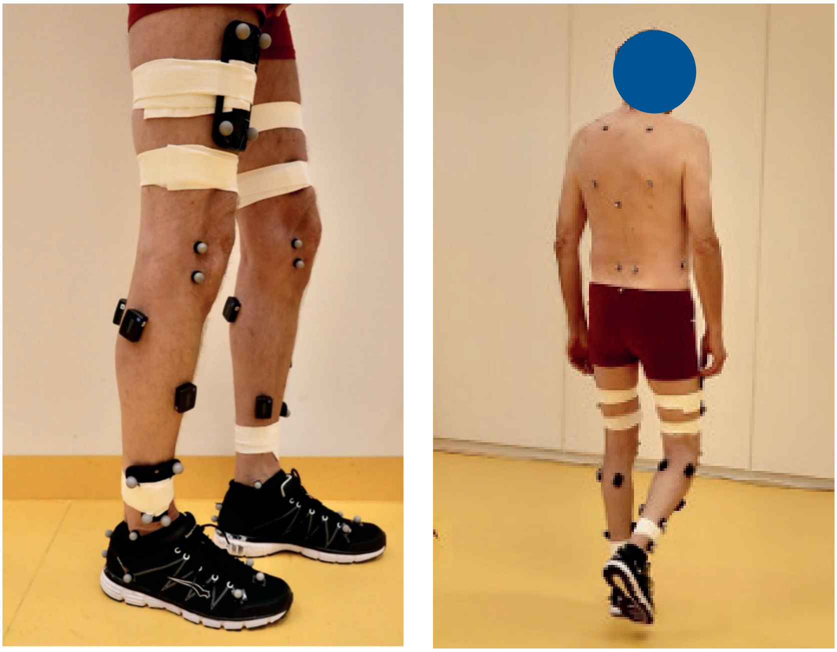

FIGURE 23: The EMG sensor application of the shank muscles: tibialis anterior (AT), medial gastrocnemius (GM), lateral gastrocnemius (GL) and soleus (SOL) and the marker model of 6 degrees of freedom (see also figure 14 for the marker model).

EMG activity was expressed as the mean percentage of %MVIC across two stance sub-phases-initial contact to mid-stance (IC–MS) and mid-stance to toe-off (MS–TO)-for the following target muscles: TA, GM, GL, and SOL. Kinematic analysis included joint excursions, calculated as the total angular displacement (i.e., the difference between maximum and minimum joint angles) for the ankle, knee, and hip joints in the sagittal plane, and for the ankle and hip joints in the frontal plane, separately for each sub-phase. Kinetic analysis involved calculating mean joint moments (Nm/kg) for the same joints and planes, also segmented by sub-phases.

EMG sensors were applied to AT, GM, GL and SOL muscles in a standing position, as outlined in the muscle application overview (Figure 23, Table

4). Bipolar bilateral EMG signals were recorded from both the affected and unaffected sides.

The elected normalization method for EMG was MVIC where participants performed four contractions of target muscles lasting for 6 s with 60 s resting periods between trials.

At a self-selected speed, timed with TC Timer (Browner Timing System), over a 10-meter walkway with four centrally positioned force plates walking and running (averaging speed of 1.36 m/s and 2.51 m/s, respectively) were performed. Five valid foot strikes on the force plates were registered for both the right and left feet.

STUDY IV

Side-to-side differences in muscle activation, joint range of motion (ROM), joint moments, joint power, and support moment were examined during the stance phase of walking and running in 22 participants, approximately one year after an acute unilateral ATR.

The same methodological approach as in Study III was used for kinetic and kinematic data collection, including the use of force plates and segmentation of the stance phase into sub-phases for the EMG calculations. EMG signals were normalized according to the same principles as in Study III (Figure 10). EMG activity was recorded using a full-surface electrode set-up applied to the TA, GM, GL, SOL, VM, GM, EO and ES (Figure 24, table 4).

Kinetic variables included sagittal plane joint moments (Nm/kg) and joint power (W/kg) obtained at the discrete time events of IC, MS, and TO respectively.

Joint moments were further used to calculate the total support moment (SM) for each limb [104, 105]. SM was defined as the sum of the sagittal plane extensor moments obtained at the ankle, knee, and hip respectively [Eq.1], to provide a measure of total limb support and interjoint coordination and providing insight into compensatory movement strategies that are not apparent when analyzing single joint moments in isolation.

$SM=-(M_{ankle}+M_{knee}+M_{hip})$ [Eq.1]

A B

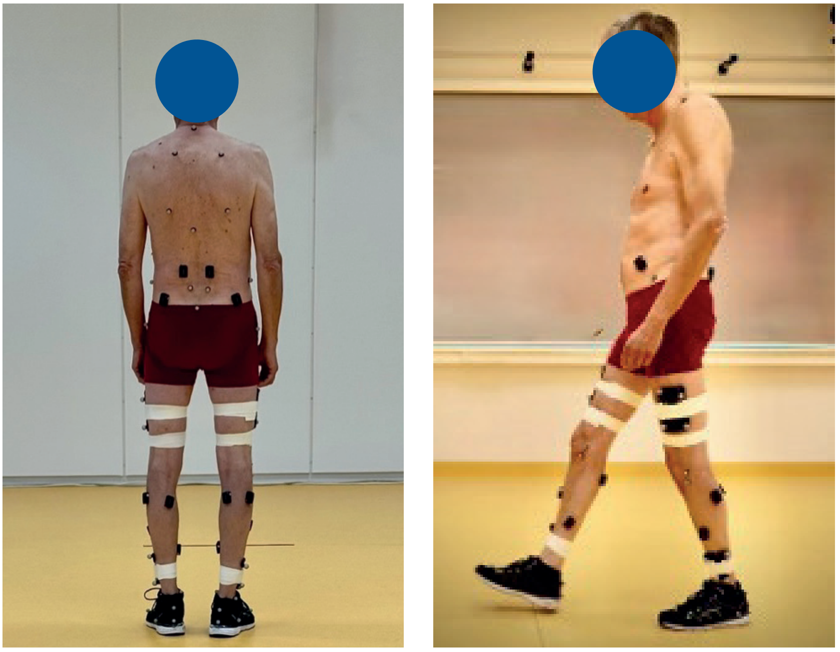

FIGURE 24. The full EMG application of the target muscles; tibialis anterior (TA), medial gastrocnemius (GM), lateral gastrocnemius (LG), soleus (SOL), vastus medialis (VM), medial gluteus (MG), external oblique (EO) and erector spinae (ES) and the marker model application of the 6 degrees of freedom (see also figure 14 for the marker model).

The signal processing in Studies III and IV were identical. EMG data was filtered by a digital high pass filter of 5 Hz, followed by a 4th order Butterworth filter in Visual 3D, v. 2025.01.1, HAS-Motion, (*Kingston, Ontario*, Canada). The data was then linearly detrended and smoothed using a centered 30-ms root mean square (RMS) window to compute the linear envelope (Figure 12). The motion capture data was processed using a 4th-order bidirectional Butterworth filter with a low-pass cut-off at 10 Hz.

In Studies III and IV, the EQ-5D and Physical Activity Scale (PAS) were included as PROMs to provide complementary information on participants' perceived health-related quality of life and physical activity.

EQ-5D

The EQ-5D [106] is a standardized instrument used to measure self-reported health-related quality of life. It shows good construct and predictive validity, along with responsiveness, supporting its use for assessing health status in patients who have sustained traumatic limb injuries [107].

EQ-5D includes five dimensions: mobility, self-care, usual activities, pain or discomfort, and anxiety or depression. Each dimension is assessed using five levels: no problems, slight problems, moderate problems, severe problems, and extreme problems. Patients indicate their current health status by selecting the statement that best describes their condition in each dimension. Each selection is represented by a single-digit score, and the five scores are combined to form a five-digit code. The code corresponds to a weighted score that summarizes an individual's overall health status. Scores range from -0.43 to 1.00, where 1.00 indicates perfect health and -0.43 reflects a health state considered worse than death.

PHYSICAL ACTIVITY SCALE

The Physical Activity Scale (PAS), [108] is a self-reported questionnaire designed to assess the level of physical activity. Initially developed to compare physiological outcomes between middle-aged individuals and athletes of the same age, the PAS uses a six-point ordinal scale, where a score of 1 indicates minimal physical activity and a score of 6 represents engagement in heavy physical activity. Although PAS has not been specifically validated for individuals after an ATR, patients who have sustained an ATR are often similar in terms of age ranges as those originally studied. The PAS provides a general estimate of activity levels, which may be useful for evaluating functional recovery or physical capacity in clinical and research setting.

STATISTICAL ANALYSIS

Each of the four studies in this thesis employed statistical approaches tailored to their specific research aims and data characteristics (for overview, see Table 6). Together, these diverse statistical approaches-frequentist, Bayesian, and time-series-provided complementary insights. They enabled both structured group comparisons and more nuanced, temporally and probabilistically informed interpretations of neuromechanical coordination following an ATR.

TABLE 6. Statistical Procedures and Software Applications utilized in Studies I-IV.

Statistical method and data programs	Study I	Study II	Study III	Study IV
Wilcoxon signed rank test	✓	✓		✓
Multivariant normal Bayesian model			✓	
Coefficient of variance (CV)			✓	
Statistical parametric mapping (SPM)				✓
Coefficient of quartile variance				✓
SPSS v. 24	✓			
SPSS v. 28		✓		
SPPS v. 30			✓	✓
R Foundation for Statistical Computing v. 4. 5. 0 with brms package v.2.21.0			✓	
Matlab (Math works, Natick, USA)				✓
G*Power v 3.1.9.7			✓	✓

Studies I and II used conventional frequentist analyses-including repeated measures ANOVA and non-parametric tests-to evaluate group-level differences in EMG amplitude across foot postures, step phases, and test positions. These methods were considered appropriate for the controlled experimental designs but were limited in their ability to model temporal dynamics or fully account for inter-individual variability.

STUDY I:

To compare MVIC EMG activity between standing and supine body positions across all target muscles, the Wilcoxon signed-rank test was used, as most of the extracted data were not normally distributed according to the Shapiro-Wilk test. Variability in MVIC EMG amplitude was assessed by examining median values and quartile distributions for each position, along with identification of outliers and extreme values. All statistical analyses were performed using SPSS, version 24.

STUDY II:

Mean EMG amplitudes (%MVIC) were compared across neutral, pronated, and supinated stance positions for given target muscles on both the right and left sides using Wilcoxon's signed-rank test ($p<0.05$, two-tailed), as the Shapiro-Wilk test revealed that the majority of the data deviated significantly from a normal distribution. The neutral stance served as the baseline condition for these comparisons. EMG amplitude for each muscle was calculated as the mean of four trials and normalized to the mean peak amplitude recorded during MVIC, expressed as a percentage of MVIC. All statistical analyses were conducted using IBM SPSS Statistics, version 28.

STUDY III:

Study III applied a multivariate Bayesian modeling framework to assess interlimb differences in terms of kinematics, kinetics, and EMG patterns during gait [109]. This approach enabled simultaneous estimation of multiple outcomes and their correlations, while quantifying uncertainty through posterior distributions. The use of 95% credible intervals allowed for probabilistic interpretation of the magnitude and direction of observed differences, aligning with the broader interpretive focus on variability and individual adaptation. This was particularly advantageous given the moderate sample size and heterogeneity in neuromuscular responses across participants.

Based on the results from a previous study evaluating biomechanical variables during walking, running, and hopping after an acute ATR [(74)], it was estimated that a sample size of 38 patients would be required to detect significant side-to-side differences in lower extremity biomechanics (α = 0.05, statistical power = 80%) at 12 months post-injury.

All processed data were analyzed using a multivariate normal Bayesian model to assess within-subject differences between the affected and unaffected limbs. Point estimates and 95% credible intervals (CrI) were defined as the posterior median and the 2.5th and the 97.5th percentiles of the posterior sample, respectively. In this model, posterior median and mean are not distinguishable; thus, descriptive statistics are reported as means and were generated using IBM SPSS Statistics (v. 30.0).

Initially, separate multivariate models were fitted for kinetic, kinematic, and electromyographic (EMG) data respectively, and then analyzed independently for walking and running. Complete case data were used for all models. As a sensitivity analysis, each outcome variable was also modeled separately using univariate linear models that included only an intercept and standard deviation.

Six models were implemented in R (R Foundation for Statistical Computing), (version 4.5.0), [(110)] using the BRMS package (version 2.21.0) [(111)], which provides an interface to Stan for Bayesian inference. Bayesian estimation was based on the joint posterior distribution, which allowed for quantification of uncertainty in all parameters, including average side-to-side differences, population standard deviations, and correlations between variables.

To fit the models, six Markov chain Monte Carlo (MCMC) chains were run in parallel, yielding 6,000 post–warm-up iterations. Convergence was evaluated using the improved R-hat statistic [111]. Default weakly informative priors were applied; a Student's t-distribution with 3 degrees of freedom for intercepts, a half-Student's t-distribution for standard deviations, and a LKJ (Lewandowski–Kurowicka–Joe distribution) prior for covariance matrices. These priors introduce mild regularization, slightly shrinking the posterior toward zero and thereby reducing the risk of type I and type M errors when dichotomizing credible intervals.

Quantitatively similar results can be obtained using penalized frequentist regression techniques such as ridge regression or LASSO. Model comparisons were conducted using the expected log pointwise predictive density (ELPD), [112]. Side-to-side differences are presented as posterior point estimates with corresponding 95% CrI.

In addition, the coefficient of variation (CV) was calculated as the ratio of the standard deviation to the mean to descriptively assess within-subject variability in kinematic, kinetic, and EMG parameters [1, 113].

Study IV employed Statistical Parametric Mapping (SPM) to compare continuous EMG%, joint angle (ROM), joint moment and joint power curves across the gait cycle. SPM allowed for time-resolved analysis of entire waveforms, facilitating the detection of phase-specific asymmetries that might be obscured by traditional pointwise comparisons. This method added a temporal dimension to the interpretation of segmental coordination.

An a priori statistical power analysis was performed using G*Power v 3.1.9.7 [114] to estimate the required sample size for a two-tailed paired *t*-test. With an alpha level of 0.05, a planned sample size of n = 23, and an expected medium effect size (d = 0.53), the estimated power was 0.80, indicating that the study was sufficiently powered to detect within-subject side-to-side effects. Temporal differences across the stance phase were assessed using Statistical Parametric Mapping, package

SPM1S in Matlab (MathWorks, Natick, USA). This time-continuous statistical analysis method has been comprehensively described in previous work [115, 116]. Descriptive statistics and SM calculations were conducted using SPSS Statistics v 30.0 (IBM Corp., Armonk, USA). As the data was not normally distributed, non-parametric Wilcoxon signed-rank testing was applied for paired comparisons for the SM of the affected and unaffected limbs. Both tests were implemented with an α-level of 0.05 (two-tailed).

For the processed EMG data, the coefficient of quartile variation (CQV) was used to evaluate variability, providing a robust measure of dispersion suitable for non-normally distributed data [117, 118]. This method was not applied to kinematic and kinetic variables, as the presence of negative values rendered the CQV results uninterpretable.

STUDY POPULATIONS

STUDIES I AND II:

Participants in Studies I and II (Tables 7 and 8) were recruited as a convenience sample from the municipal area of Gothenburg, either through Facebook advertisements or from the nursing and physiotherapy programs at the University of Gothenburg.

STUDIES I AND II

TABLE 7. Demographic data from Study I and II (n=12).

Age (years)	33.3.(± 9.6)
Height (cm)	173.6 (±7.8)
Weight (kg)	71.6.(±11.3)
BMI	23.5 (± 2.6)

Data are presented as mean and standard deviation (SD). BMI = body mass index.

TABLE 8. Anthropometric characteristics of the lower extremities from Study II (n=12).

Anthropometric Characteristics	Participants
Leg length R foot	89.9 (±7.3)
Leg length L foot	90.0 (±7.5)
ND R foot Mean mm (SD	5.3 (±1.3)
ND L foot	4.8 (±1.2)

Data are presented as group mean and standard deviation (±SD). R = right; L = left; ND: navicular drop measure.

STUDIES III AND IV:

Participants in Studies III (Table 9) and IV (Table 10) were recruited from the ongoing DUSTAR project (Diagnostic Ultrasonography for Sahlgrenska Academy, University of Gothenburg. Data collection was conducted approximately one year (+2 months) following the Achilles tendon injury.

A total of 37 participants were included in Study III, where EMG recording was applied to selected muscles of the shank. In Study IV, a subset of the final 22 participants underwent extended EMG recordings, which included muscles along the full biomechanical chain under investigation (Figure 25).

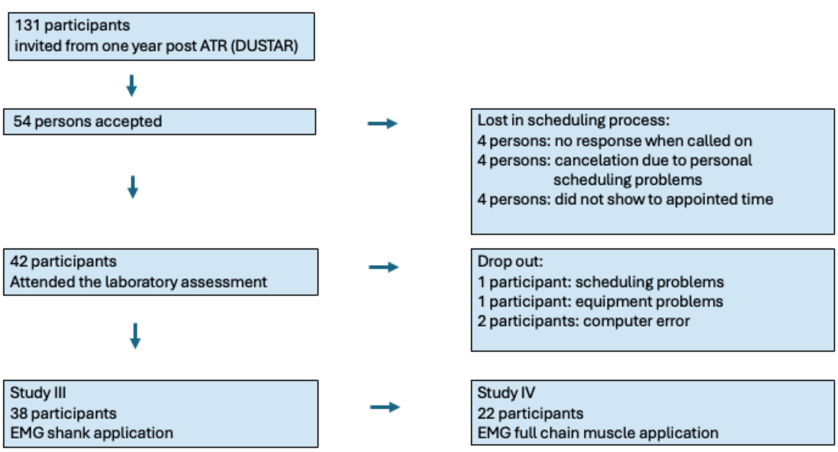

FIGURE 25. Flow chart of participant inclusion in Studies III and IV.

STUDY III

TABLE 9. Participant demographics and patient reported outcome measurements score in Study III (n=37).

Age (years)	47.4 (± 9.4)
Height (meter)	1.79 (± 0.08)
Weight (kg)	83.3 (± 16.6)
BMI	26.9 (±4.9)
PAS	4 (± 1.1)
EQ-5D	0.7 (± 0.3)

Data are presented as group mean and standard deviation (±SD). BMI = Body mass index; PAS= Physical activity scale (six-graded scale), self-reported level of physical activity (1.denoting extremely low physical activity and 6.denoting strenuous physical activity); EQ-5D self-reported score health-related quality of life (range 1- (-0.43), 1 = perfect health. -0.43 = worse than death.

STUDY IV

TABLE 10. Participant demographics and patient reported outcome measurements score in Study IV (n=22).

Age (years)	48.0 (±10.9)	
Height (meter)	1.79 (± 0.08)	
Weight (kg)	83.3 (± 16.6)	
BMI	26.7 (±3.3)	
PAS	4.1 (± 1.2)	
EQ-5D	0.8 (± 0.2)	
Treatment:	Surgery: 14 participants	No surgery: 8 participants

Data are presented as group mean and standard deviation (±SD). BMI = Body mass index; PAS = Physical activity scale (six-graded scale), self-reported level of physical activity (1 denoting extremely low physical activity and 6 denoting strenuous physical activity); EQ-5D self-reported score health-related quality of life (range 1- (-0.43), 1= perfect health. -0.43 = worse than death.

7. Summary of Results

"

Say, who is this walking man?

JAMES TAYLOR

STUDY I

Participants were recruited as a convenience sample from the Gothenburg area to compare two standardized test positions-supine and standing-for EMG normalization using MVIC. The aim was to determine whether body posture affects MVIC values for selected muscles in the trunk and lower extremities.

No systematic differences in terms of EMG amplitudes were found when comparing the supine and standing positions for MVIC testing, suggesting that both postures are equally effective for normalizing muscle activity when it comes to group-level analyses (Figure 26).

Bars show mean values; error bars indicate ±1 standard deviation. Muscles: TA = tibialis anterior, GlM = gluteus medius, AddL = adductor longus, RA = rectus abdominis, EO = external oblique, IO/TrA = internal oblique/transversus abdominis.

Substantial inter-individual variability was observed in muscle activation levels in both positions, indicating that normalization outcomes may differ across individuals regardless of the test posture (Figure 27).

Muscles: tibialis anterior (TA), gluteus medius (GM), adductor longus (AddL), rectus abdominis (RA), external oblique (EO), and internal oblique/transversus abdominis (IO/TrA). R = right, L = left.

Outliers are shown as circles (O); extreme values are marked with asterisks (*). Muscle activation did not differ significantly between the two positions (Figure 27).

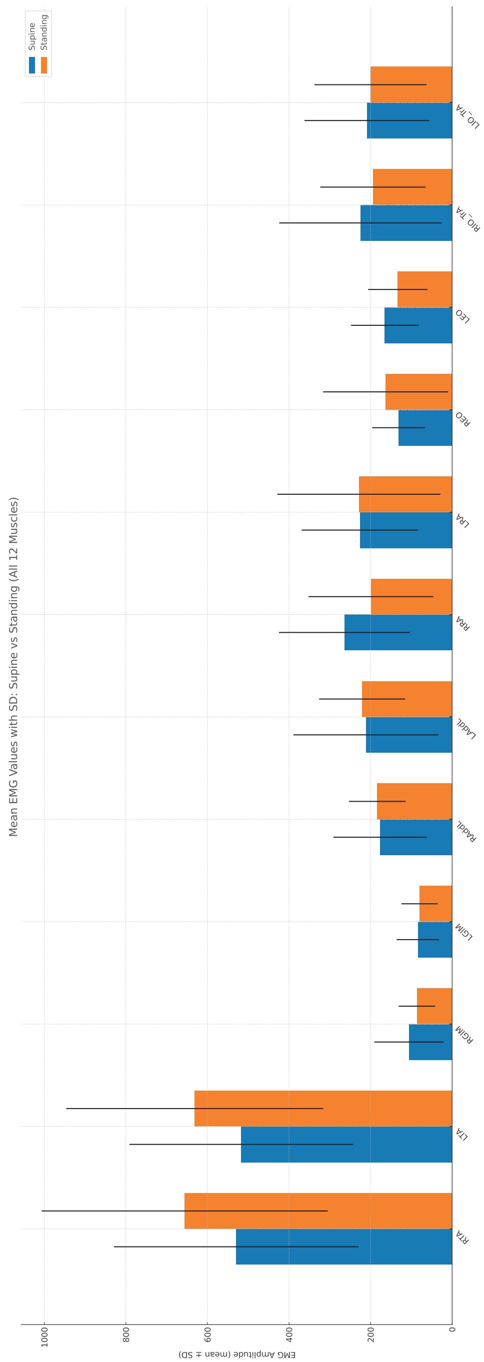

FIGURE 26. EMG amplitudes (μV) during maximal voluntary isometric contractions (MVIC) of selected muscles in supine (blue) and standing (orange) postures.

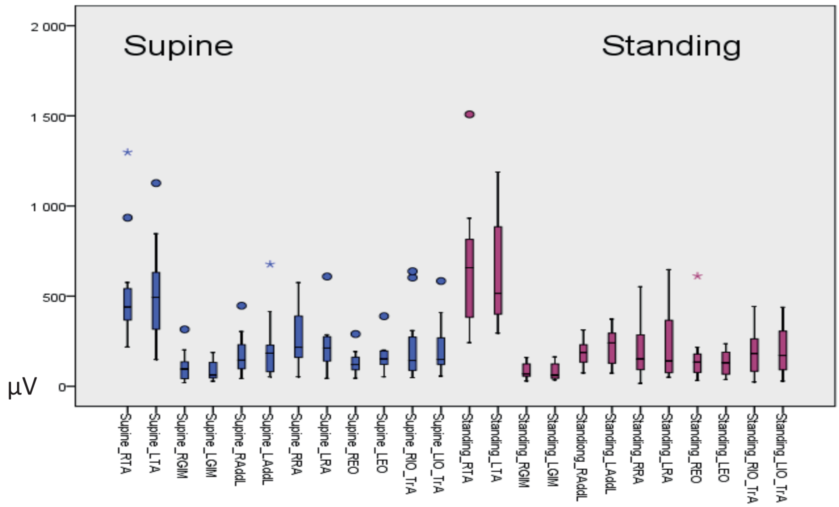

FIGURE 27. Inter-individual variability (median ± interquartile range) in maximum EMG amplitude (µV) for selected lower-limb and trunk muscles during maximal voluntary isometric contractions (MVIC) performed in supine or standing positions.

STUDY II

This study investigated how foot kinematics-specifically supinated and pronated positions compared with neutral affect muscle activation in selected lower extremity and trunk muscles. The foot position on the ipsilateral stance foot was manipulated, while the contralateral limb performed the step during a standardized vertical step maneuver. Participants were recruited from the same convenience sample as in Study I.

Study II demonstrated that foot configuration had limited systematic influence on neuromuscular activation during a single-step stair ascent, although some significant differences were observed in individual muscles. Compared with the neutral baseline, foot supination and pronation resulted in significant changes in a few muscles but showed no systematic influence on trunk activation, whereas the lower extremity displayed more significant differences, yet these did not reflect a substantial or systematic overall influence. Moreover, the high inter-individual variability in EMG data indicates that within-subject motor variability plays a crucial role in maintaining postural stability and task performance, particularly during functional activities exposed to perturbations.

Right foot stance:

Ipsilateral: In the pronated position, the right EO (p=0.010) and TA (p=0.023) showed a higher activity than in neutral. In the supinated position, the right TA (p=0.004), PE (p=0.019), and SOL (p=0.005) showed a higher activity than in the neutral position.

***Contralateral*:** In the pronated position, the left MU showed a higher activity than in the neutral position (p=0.034), (Figure 28).

Left foot stance:

Ipsilateral: In the supinated position, the left EO showed a higher activity than in the neutral position (p=0.023), while the left AddL (p=0.041) and SOL (p=0.050) showed lower activity than in the neutral position.

Contralateral: In the pronated position, the right MU showed a higher activity than in the neutral position (p=0.033), and in the supinated position, the right IO/TrA showed a higher activity than in the neutral position (p=0.023), (Figure 29).

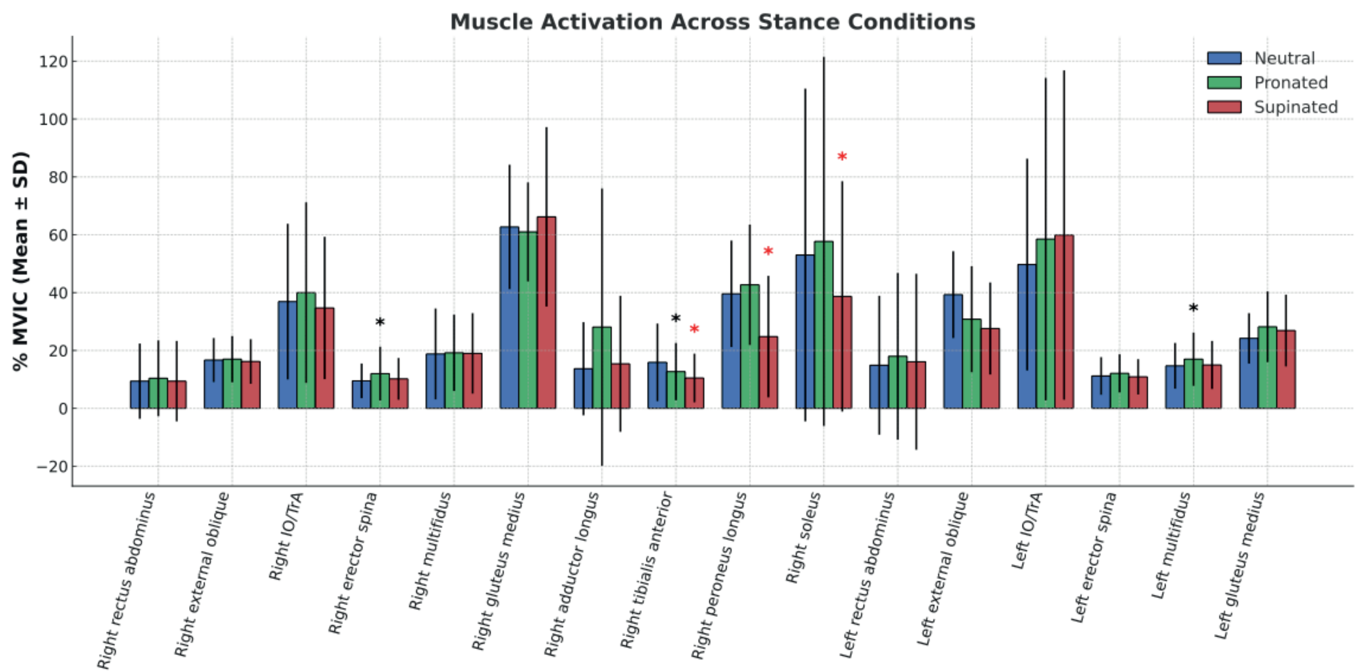

FIGURE 28. Muscle activation assessed as mean EMG activity (SD) for the stance limb (right trunk and lower limb) and contralateral limb (left trunk and left gluteus medius) muscles during a standardized single-step stair ascent. IO/TrA = muscle internal oblique/transversus abdominis. The stance foot (right side) was placed in pronated, supinated, or neutral position (base line). Muscles showing significant differences in activation compared with the neutral foot position: * pronated-right erector spinae, right tibialis anterior, left multifidus; * supinated-right tibialis anterior, right soleus, n = 12.

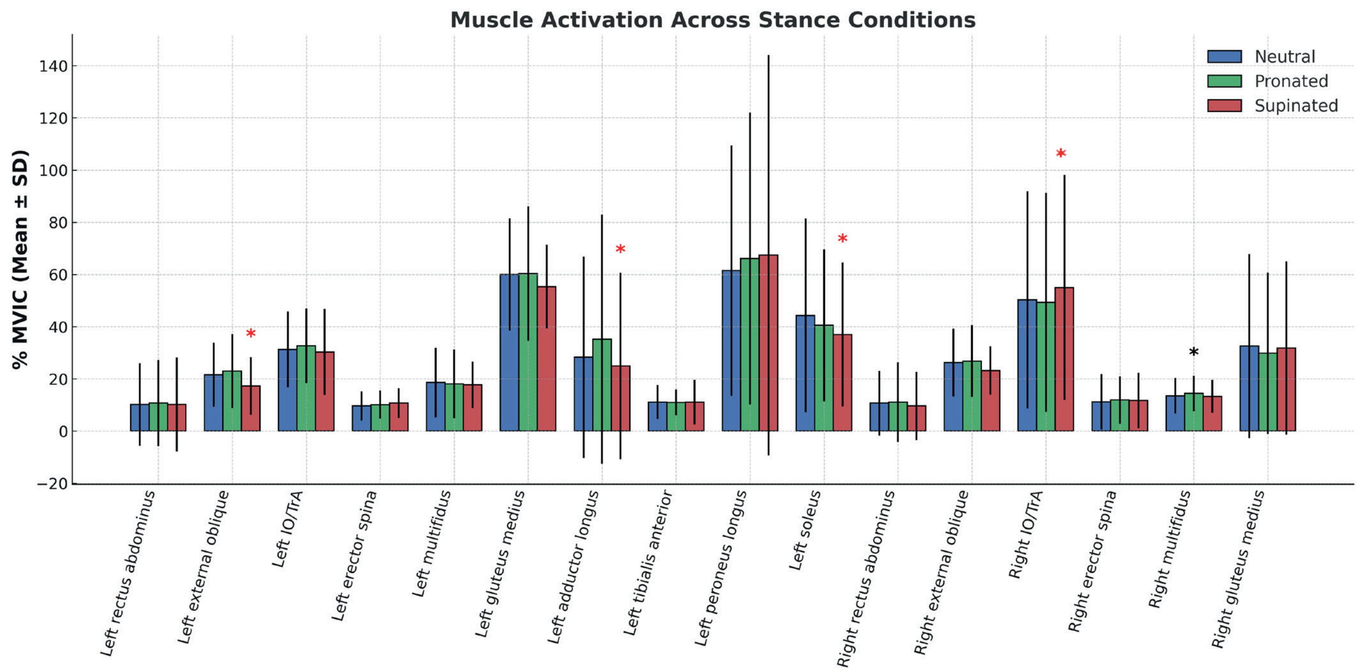

FIGURE 29. Study II, left stance. Muscle activation assessed as mean EMG activity (SD) for the stance limb (left trunk and lower limb) and contralateral limb (right trunk and right gluteus medius) muscles during a standardized single-step stair ascent. IO/TrA = muscle internal oblique/transversus abdominis. The stance foot (left side) was placed in pronated, supinated, or neutral position (baseline). Muscles showing significant differences in activation compared with the neutral foot position: * pronated-right multifidus, * supinated-left external oblique, left adductor longus, left soleus, n = 12.

STUDY III

A total of 37 participants were recruited from the ongoing DUSTAR project (Diagnostic Ultrasonography for the Choice of Treatment of Acute Achilles Tendon Rupture). Data collection took place approximately one year (+2 months) after the ATR. The study aimed to investigate limb-to-limb differences for lower extremity muscle activation, joint kinematics, and kinetics during the stance phase of walking and running. EMG was applied to selected shank muscles, while OTS was used to assess both kinematic and kinetic parameters in the affected as well as the unaffected limbs.

Posterior point estimates indicated biomechanical alterations within the shank complex, with the strongest evidence for differences in lower-limb muscle activation and sagittal-plane kinematics. The kinematic data indicated relatively low inter-individual variability in ankle, knee, and ROM. In contrast, joint moments and EMG measures exhibited substantially greater inter-individual variability, suggesting more diverse neuromuscular and kinetic adaptation strategies across participants. The results below summarize these comparisons for two distinct periods of the stance phase: IC–MS and MS–TO.

When assessing muscle activation across the stance-phase periods during walking and running, all examined muscles showed higher EMG activity in running, with the largest increments observed in the triceps surae muscle group (Figure 30).

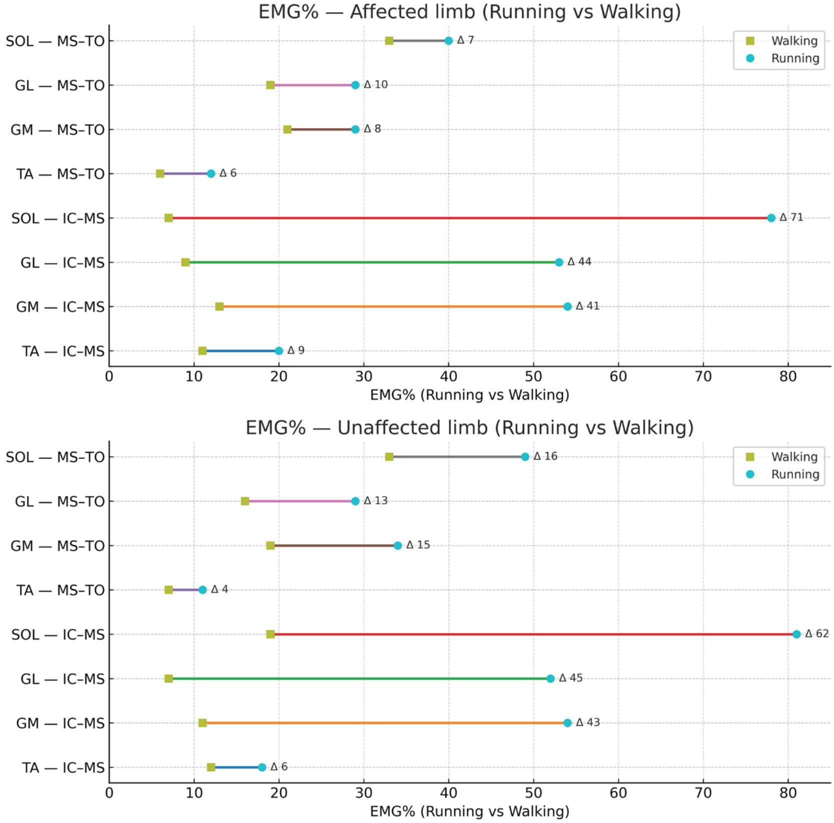

FIGURE 30. EMG% (%MVIC) recorded during walking and running by muscle and stance periods. Dumbbell plots show Walking and Running values for each stance period, with connecting lines and Δ labels indicating the difference (Walking − Running) in EMG %. Top panel: affected limb; bottom panel: unaffected limb.

Phases: IC–MS = initial contact to midstance; MS–TO = midstance to toeoff. Muscles: TA = tibialis anterior; GM = gluteus medius; GL = gastrocnemius lateralis; SOL = soleus.

Inter-limb differences were observed in Study III. During walking, posterior point estimates indicated greater LG activation in the affected limb during IC–MS, with a mean difference of 2.1 EMG% (95% CrI: 0.5 to 3.7). In MS–TO, the affected limb also showed higher activation in both

MG (3.4 EMG%, 95% CrI: 0.6 to 6.3) and LG (4.8 EMG%, 95% CrI: 2.3 to 7.6) compared to the unaffected side. Ankle sagittal ROM was reduced in the affected limb, with a posterior mean difference of -1.8° (95% CrI: -2.8° to -0.7°), (Figure 31).

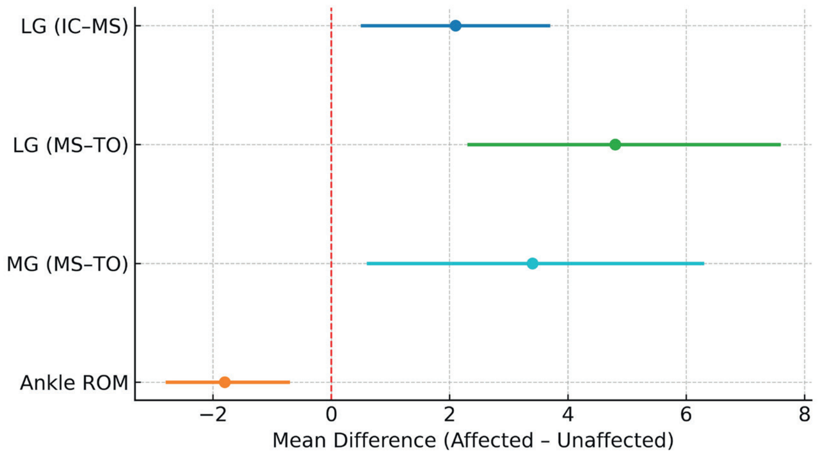

FIGURE 31. Walking. Posterior point estimates of between-limb differences during walking one year after Achilles tendon rupture for EMG % activation (GL= Gastrocnemius lateralis, GM = gastrocnemius medialis) and Ankle sagittal ROM in Nm/kg. IC-MS= initial contact to midstance, MS-TO = midstance to toe off, ROM = range of motion. Circles represent posterior means; horizontal lines represent 95% credible intervals (CrI).

Annotations indicate posterior mean (95% CrI) and posterior probability of direction. Positive values indicate greater activation or excursion in the affected limb, negative values indicate reductions.

During running, ankle sagittal ROM was reduced in MS -TO on the affected side, with a mean difference of-4.1° (95% CrI: -5.8 to -2.5), along side a lower ankle sagittal moment (0.06 Nm/kg, 95% CrI: 0.01 to 0.11) compared with the unaffected side (Figure 32).

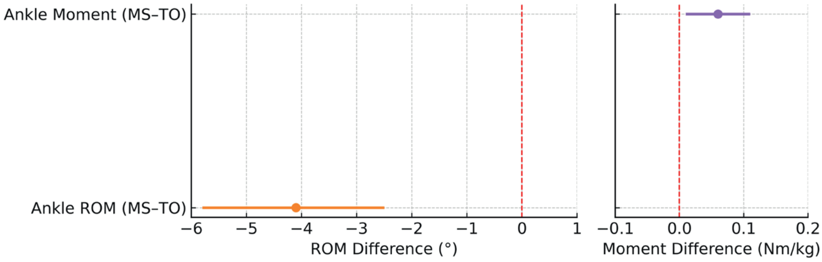

FIGURE 32. Running. Posterior estimates of between-limb differences during running one year after Achilles tendon rupture for ankle sagittal ROM in degrees and ankle sagittal moment in Nm/kg. ROM = range of motion, MS-TO = midstance to toe off. Dots represent posterior means; horizontal lines represent 95% CrI. Annotations indicate posterior mean (95% CrI) and posterior probability of direction. Positive values indicate greater ROM or moment in the affected limb, negative values indicate reductions.

For variables in the frontal plane, no differences between the affected and unaffected limbs were observed, neither during walking nor running.

Variability in joint moments, EMG activity, and ROM during walking and running was quantified using the coefficient of variation (CV). For walking, variability was lowest in ankle, knee, and hip ROM (CV: 9-38%), higher in EMG activity (CV: 38-92%), and widest in joint moments (CV: 9-900%), (z).

During running, kinematic variability was greater (CV: 16-57%), EMG variability remained high (CV: 46-136%), and joint moments again showed the largest distribution (CV: 17-660%), (Figure 33). The very high CV values for joint moments reflect the mathematical definition of CV, where the standard deviation (nominator) is divided by the mean (denominator). When mean joint moments were close to zero, even modest absolute variability resulted in comparatively large CV values.

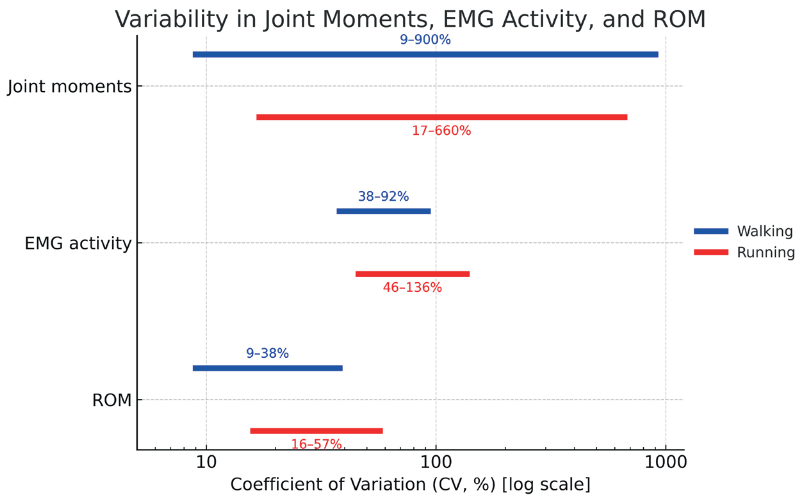

FIGURE 33. Inter-individual variability (coefficient of variation, CV%) in range of motion (ROM), electromyography (EMG) activity, and joint moments during walking and running one year after an Achilles tendon rupture (ATR).

Mean EMG activation profiles with variability are presented for the TA, MG, LG, and SOL during the stance phase of walking and running (Figures 34-37).

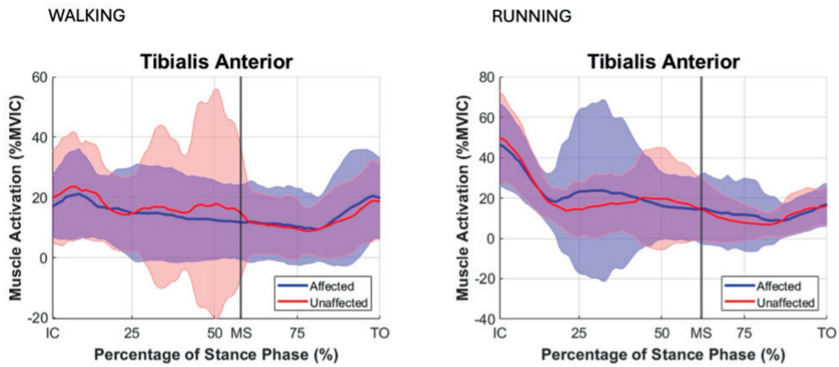

FIGURE 34. Muscle activation configuration of Tibialis anterior (see capture Figure 37).

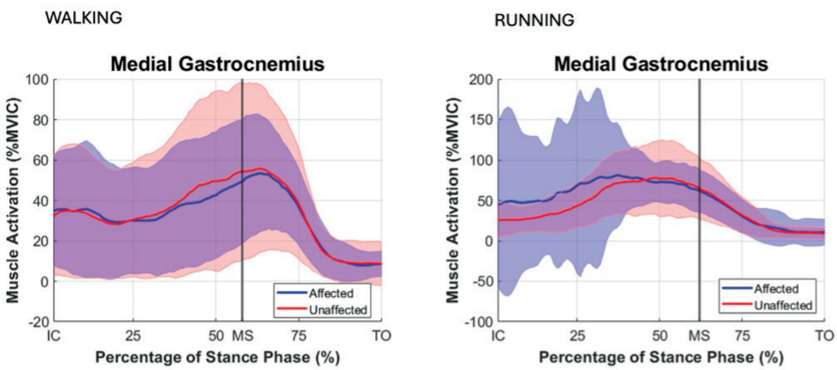

FIGURE 35. Muscle activation configuration of Medial gastrocnemius (see capture Figure 37).

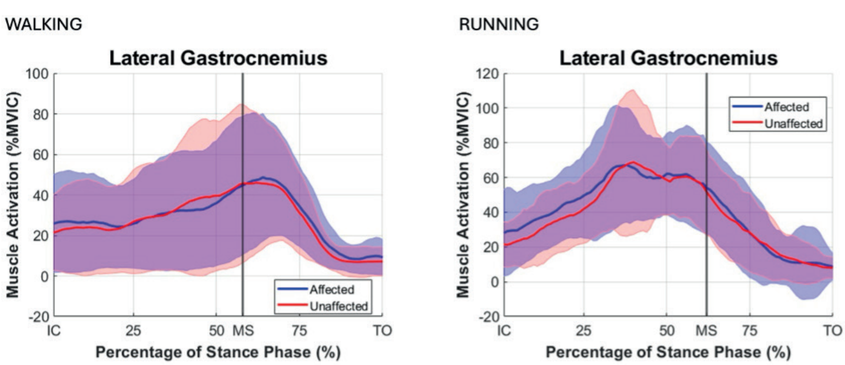

FIGURE 36. Muscle activation configuration of Lateral gatrsocnemius (see capture Figure 37).

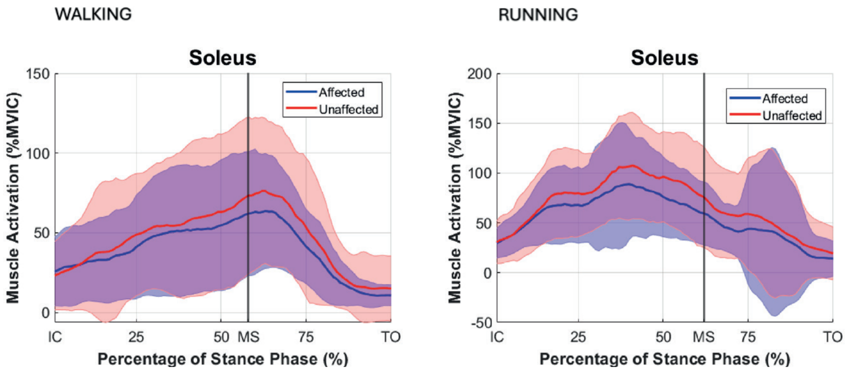

FIGURE 37. Muscle activation configuration of Soleus. Mean normalized EMG activity (%MVIC) with inter-individual variability (±1 SD) for the affected (blue) and unaffected (red) limbs. For stance phase in tibialis anterior (TA), medial gastrocnemius (MG), lateral gastrocnemius (LG), and soleus (SOL). Solid lines represent mean normalized EMG (%) across participants; shaded areas indicate inter-individual variability (±1 SD). The x-axis shows the stance phase of gait, from initial contact (IC) through midstance (MS) to toe-off (TO).

STUDY IV

A subset of 22 participants from the same cohort as Study III were included in Study IV. This study aimed to investigate side-to-side differences in muscle activation, ROM, joint power, joint moments, and support moments during walking and running one year after an unilateral ATR. The analysis focused on segmental adaptations within the ankle–knee–hip complex and their distribution along the biomechanical chain up to the trunk. Bilateral EMG recordings were collected from eight muscles, as well as kinematic and kinetic measurements to assess coordination across the lower limb and trunk.

EMG% values were consistently higher during running compared with walking, with the largest differences observed in the triceps surae (gastrocnemius medialis, gastrocnemius lateralis, and soleus) during early stance (Figures 38, 39).

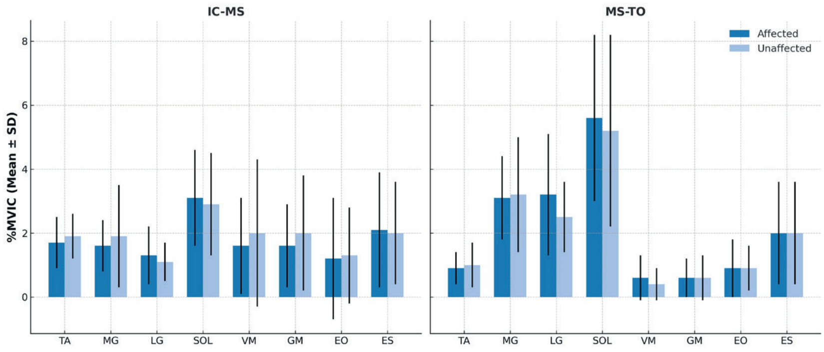

FIGURE 38. Mean (±SD) normalized EMG activity during Walking across eight muscles: tibialis anterior (TA), medial gastrocnemius (MG), lateral gastrocnemius (LG), soleus (SOL), vastus medialis (VM), gluteus medius (GM), external oblique (EO), and erector spinae (ES). Results are shown separately for early stance (IC–MS) and late stance (MS–TO), comparing the affected limb (dark blue) to the unaffected limb (light blue).

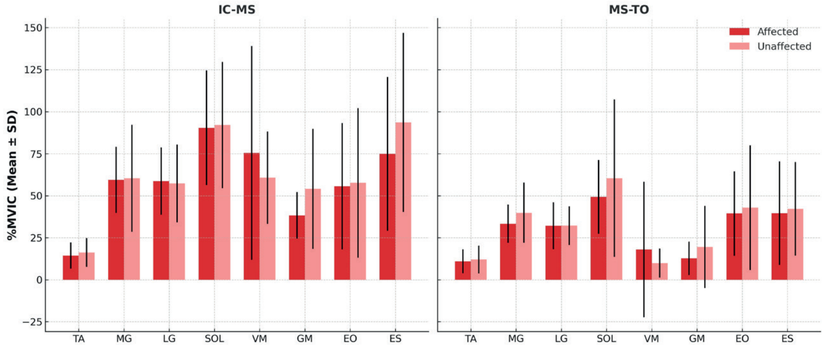

FIGURE 39. Mean (±SD) normalized EMG activity during Running across eight muscles: tibialis anterior (TA), medial gastrocnemius (MG), lateral gastrocnemius (LG), soleus (SOL), vastus medialis (VM), gluteus medius (GM), external oblique (EO), and erector spinae (ES). Results are shown separately for early stance (IC–MS) and late stance (MS–TO), comparing the affected limb (dark red) to the unaffected limb (light red).

Side-to-side differences observed in Study IV (Figure 40). During walking the affected limb showed significantly higher LG activation near the end of stance (~88–90%) compared with the unaffected limb (p=0.05), (Figure 41). Ankle ROM was significantly reduced in the affected limb during early stance (~17–20%, p=0.04), (Figure 42), and ankle joint power was lower near late stance (~88–90%, p=0.05), (Figure 44), reduced knee power was also observed in early stance (~8–10%, p=0.032), (Figure 45). Total support moment (SM) was significantly lower in the affected limb at both initial contact (IC) and toe-off (TO), (p=0.001 for both), (Figure 46).

Running: No significant side-to-side differences were observed in muscle activation or joint power. However, ankle ROM was reduced in the affected limb during late stance (~77–100%, p=0.012), (Figure 43), and total SM was significantly lower at TO (p=0.001), (Figure 47).

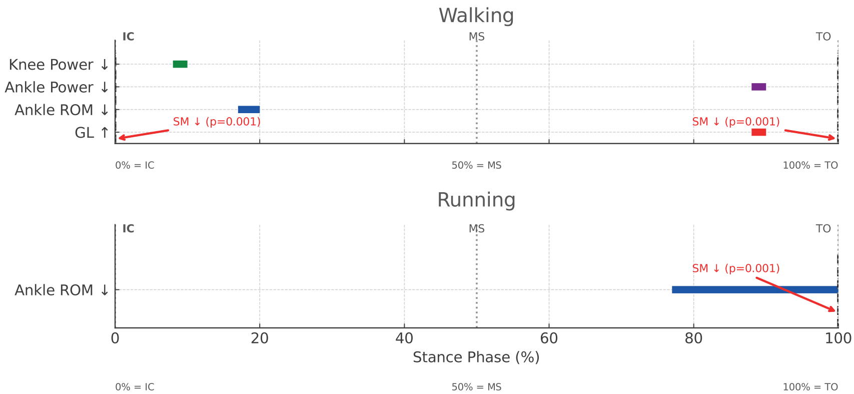

FIGURE 40. Overview of the side-to-side differences in Study IV for joint kinematics, kinetics, and muscle activation one year after an Achilles tendon rupture (ATR). Sagittal plane joint kinematics, kinetics, and muscle activation during walking (top panel) and running (bottom panel). Vertical dashed lines indicate stance events (0% = initial contact [IC], 50% = midstance [MS], 100% = toe-off [TO]). Red arrows mark phases where total support moment (SM) was significantly reduced (*p* = 0.001).

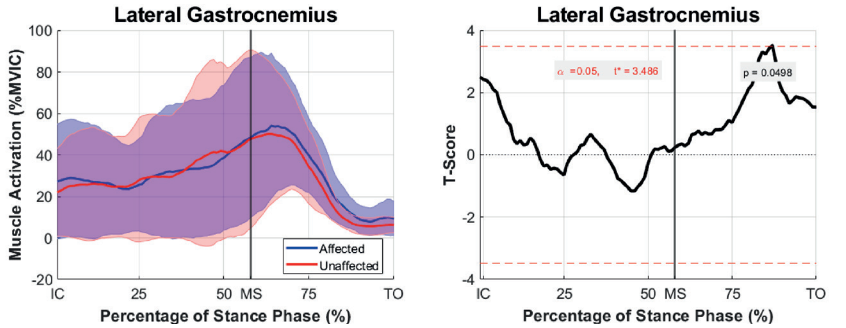

FIGURE 41. Lateral gastrocnemius (GL) muscle activity during the stance phase of walking one year after Achilles tendon rupture. *Left panel*: Mean normalized EMG activity (%MVIC) with inter-individual variability (±1 SD) for the affected (blue) and unaffected (red) limbs. *Right panel*: One-dimensional paired t-test (SPM) comparing affected and unaffected sides. The black curve represents the t-statistic across stance, with the critical threshold (α = 0.05) shown by the dashed red line. Significant difference between affected and unaffected limbs was seen in later part of stance.

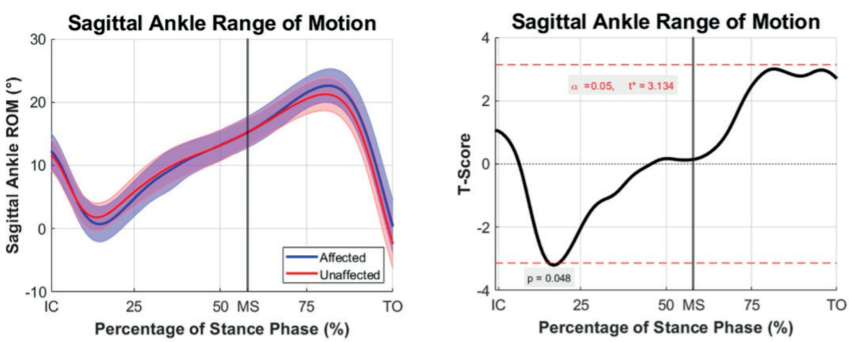

FIGURE 42. Ankle sagittal ROM measured during the stance phase of walking one year after an Achilles tendon rupture. *Left panel*: Mean ankle ROM in degrees (°) with inter-individual variability (±1 SD) for the affected (blue) and unaffected (red) limbs. *Right panel*: One-dimensional paired t-test comparing affected and unaffected sides. The black curve represents the t-statistic across stance, with the critical threshold (α = 0.05) shown by the dashed red line. Significant differences between affected and unaffected limbs were seen in the early part of stance.

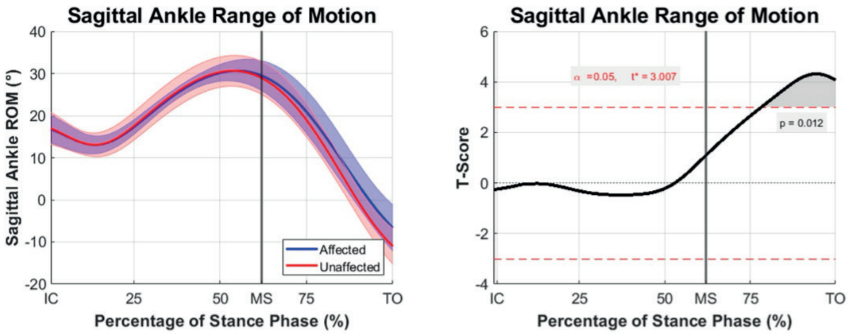

FIGURE 43. Ankle sagittal ROM measured during the stance phase of running one year after Achilles tendon rupture. *Left panel*: Mean ankle ROM in degrees (°) with inter-individual variability (±1 SD) for the affected (blue) and unaffected (red) limbs. *Right panel*: One-dimensional paired t-test (SPM) comparing affected and unaffected sides. The black curve represents the t-statistic across stance, with the critical threshold (α = 0.05) shown by the dashed red line. Significant differences between affected and unaffcetd limbs were observed in the late part of stance.

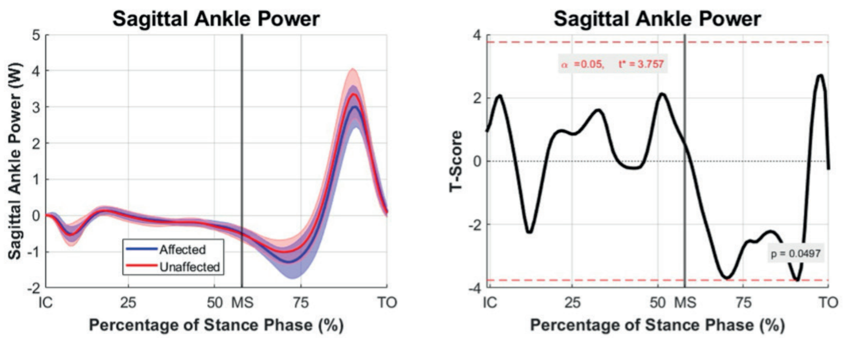

FIGURE 44. Sagittal ankle joint power measured during the stance phase of walking one year after Achilles tendon rupture. *Left panel*: Mean ankle power (W/kg) with inter-individual variability (±1 SD) for the affected (blue) and unaffected (red) limbs. *Right panel*: One-dimensional paired t-test (SPM) comparing affected and unaffected sides. The black curve represents the t-statistic across stance, with the red dashed line marking the significance threshold (α = 0.05). Significant differences between affected and unaffected limbs were seen in the later part of stance.

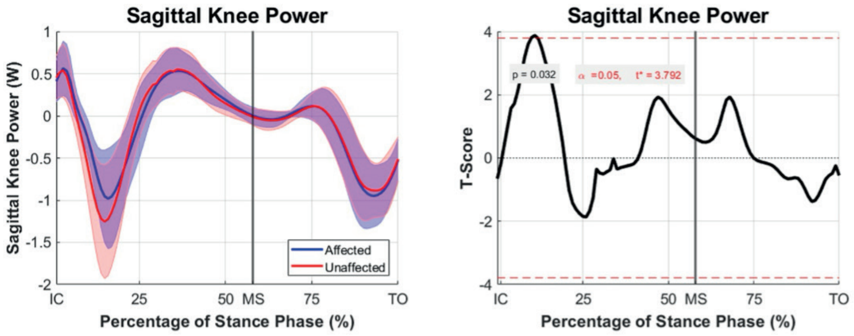

FIGURE 45. Sagittal knee joint power generation during the stance phase of walking one year after Achilles tendon rupture. *Left panel*: Mean knee power (W/kg) with inter-individual variability (±1 SD) for the affected (blue) and unaffected (red) limbs. *Right panel*: One-dimensional paired t-test (SPM) comparing affected and unaffected sides. The black curve represents the t-statistic across stance, with the red dashed line marking the significance threshold (α = 0.05). Significant differences between affected and unaffected limbs were seen in the early part of stance.

Total support moment (SM):

Although individual joint moments calculations did not differ significantly between affected and unaffected limbs, the calculated support moments did. In walking, significant differences were found at IC (p<0.001) and TO (p<0.001), (Figure 46), and in running at TO (p<0.001), (Figure 47), with lower values in the affected limb compared with the unaffected.

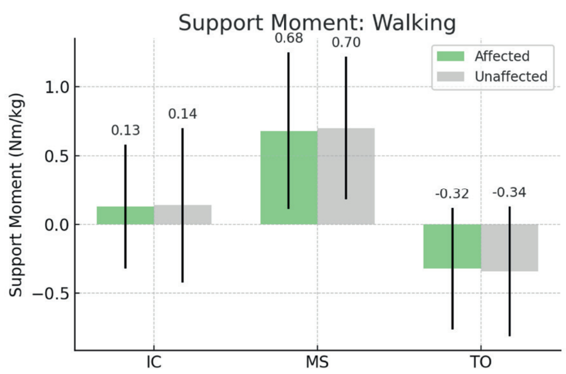

FIGURE 46. Support moment (Nm/kg) during walking for affected (green) and unaffected (grey) limbs (n = 22). Bars represent mean values at initial contact (IC), midstance (MS), and toe-off (TO). Error bars indicate SD. Asterisks (*) mark significant differences between limbs (p<0.05).

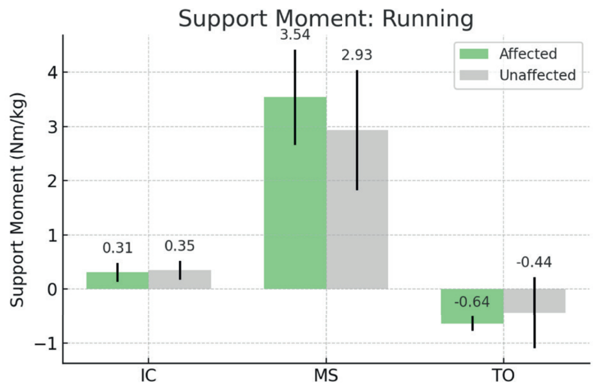

FIGURE 47. Support moment (Nm/kg) during running for affected (green) and unaffected (grey) limbs (n = 22). Bars represent mean values at initial contact (IC), midstance (MS), and toe-off (TO). Error bars indicate SD. Asterisks (*) mark significant differences between limbs ($p<0.05$).

Motor variability

The coefficient of quartile variation (CQV) revealed distinct patterns of inter-individual variability across muscles, stance phases, and limbs during both walking and running (Figures 48, 49). Across-gait modes, ankle plantar flexors displayed more consistent activation profiles, while proximal and trunk muscles demonstrated greater inter-individual variability.

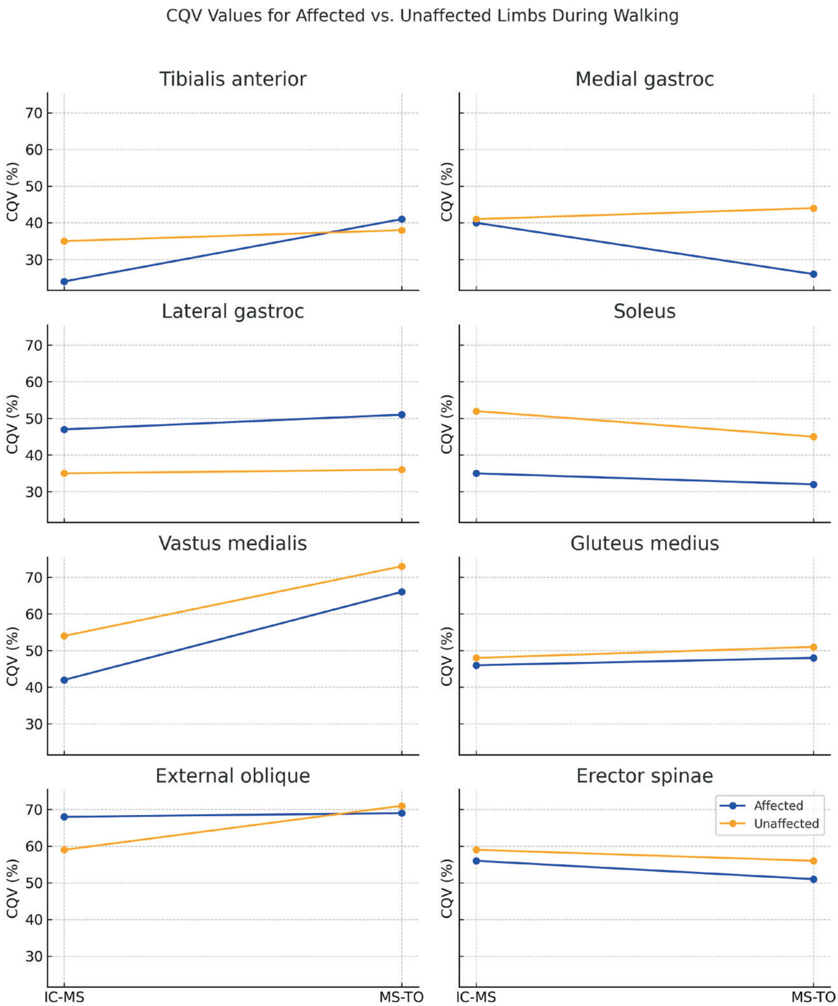

FIGURE 48. Coefficient of quartile variation (CQV, %) for affected and unaffected limbs during walking, shown separately for eight muscles: tibialis anterior, medial gastrocnemius, lateral gastrocnemius, soleus, vastus medialis, gluteus medius, external oblique, and erector spinae. CQV values are presented for early stance (IC–MS = initial contact to midstance) and late stance (MS–TO = midstance to toe-off). Blue lines represent the affected limb and yellow lines the unaffected limb.

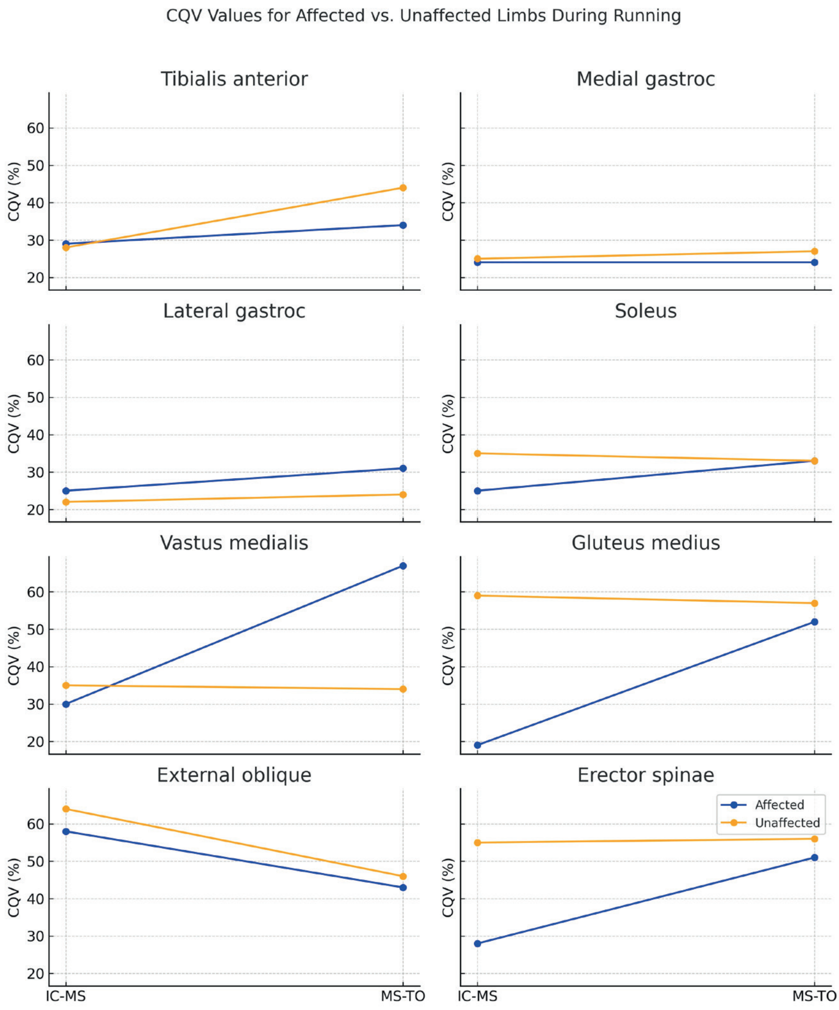

FIGURE 49. Coefficient of quartile variation (CQV, %) for affected and unaffected limbs during running, shown separately for eight muscles: tibialis anterior, medial gastrocnemius, lateral gastrocnemius, soleus, vastus medialis, gluteus medius, external oblique, and erector spinae. CQV values are presented for early stance (initial contact to midstance, IC–MS) and late stance (midstance to toe-off, MS–TO). Blue lines represent the affected limb and yellow lines the unaffected limb.

8. Discussion

"

It's very hard in the beginning to understand that the whole idea is not to beat the other runners. Eventually, you learn that the competition is against the little voice inside you that wants you to quit.

GEORGE SHEEHAN

The main contributions of this thesis are first, the demonstration that EMG normalization is reliable and second, the observation that changes in foot configuration are mainly absorbed locally rather than propagated proximally. After an Achilles tendon rupture, neuromechanical adaptations remain largely localized to the lower leg. Persistent ankle deficits and occasional knee compensations occur, but overall gait patterns resemble normative kinematics.

The discussion section is organized around four identified knowledge gaps: 1) methodological, 2) methodological/biomechanical, 3) clinical, and 4) integrative (Figure 50), each of which are addressed in the following sections.

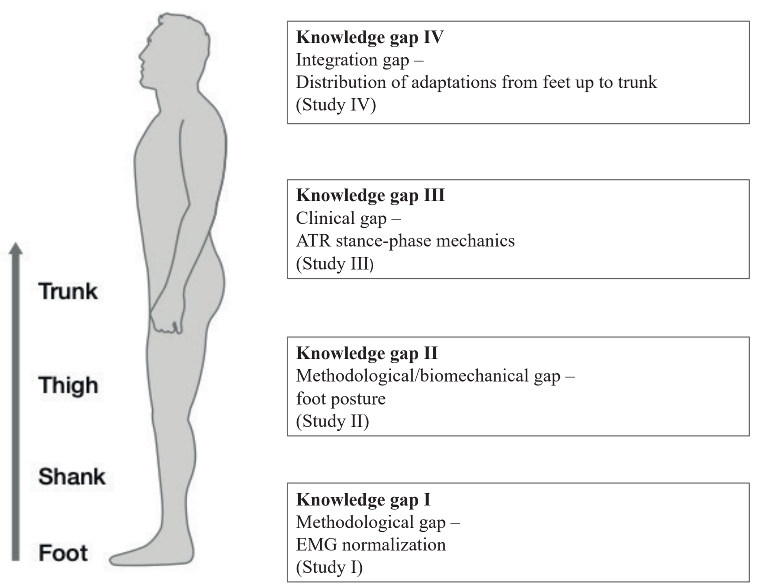

FIGURE 50. A visualization of the four knowledge gaps identified in this thesis and how they correspond to Studies I–IV. EMG = electromyography

Methodological Gap – EMG normalization (STUDY I)

Normalization of surface EMG signals to MVIC values is essential for between-participants comparability, yet differences in testing posture could influence the outcome [98]. Posture can alter muscle activation magnitude and recruitment strategy because supine positions remove load-bearing and postural stabilization demands, while upright positions engage trunk and lower-limb muscles in a more functional manner.

In Study I, upright and supine MVIC testing for lower-limb and trunk muscles yielded comparable values with showed excellent test–retest reliability (ICC = 0.80–0.90), indicating that EMG normalization is biomechanically stable and reproducible across positions, when protocols are applied consistently. Nonetheless, upright testing more closely reflects the neuromuscular demands of dynamic locomotion and may therefore provide greater ecological validity for gait-related research.

Methodological/Biomechanical Gap – Foot posture (Study II)

The biomechanical chain model proposes that distal changes, such as altered foot posture, can propagate proximally through mechanical and neural coupling [119-122]. Manipulating stance foot posture (neutral, pronation, supination) during a vertical step maneuver altered muscle activation in the lower extremity to some extent, while less effects on the trunk and no systematic influences were seen. Distal alignment changes were largely absorbed locally, with lower-leg muscles more responsive to foot kinematics than trunk muscles.

Kinematic and neuromuscular links between foot posture and proximal segments have been demonstrated [123-125]. Calcaneal eversion alters pelvic alignment through shank and thigh rotation [121, 126]. Pelvic motion changes occur with altered foot positions or insole inclination, though spinal posture remains unaffected [127, 128]. Trunk kinematics adapt to foot inclination during prolonged standing without changes in muscle thickness [129]. Foot pronation and supination modulate lower-limb muscle activity, with limited effects on trunk muscles except the erector spinae [130]. In population data, foot pronation showed an association with back pain in women, though no consistent links were found overall [131].

Evidence also suggests limited distal–proximal coupling [132-135]. No correlation was found between the foot position and pelvic or lumbar motion during standing [33]. Bilateral EMG recordings showed that extensive pronation or supination did not alter activity in trunk muscles such as latissimus dorsi, pectoralis major, or rectus abdominis [29]. No association was detected between hyperpronation and severity of nonspecific low back pain [34].

Changes in foot kinematics have been associated with overuse in gait, and foot orthoses are sometimes recommended for other than foot problems/symptoms only [136, 137], including the knee, hip, and back, although evidence is still limited. These patterns, together with indications of neuromuscular action, postulate a relationship between foot function and proximal control. In neurological terms, sensory input from plantar mechanoreceptors can affect multi-joint coordination, potentially eliciting trunk responses within 100–200 ms [138, 139].

Altered foot posture primarily affects lower-leg mechanics and muscle activity, with limited and inconsistent effects on pelvic alignment or trunk function. While neural pathways such as plantar sensory input may link distal and proximal segments, current evidence indicates that most adaptations are absorbed locally, suggesting only modest distal–proximal coupling.

Clinical Gap – ACHILLES TENDON RUPTURE stance-phase mechanics (STUDY III)

The Achilles tendon can store and release elastic energy during locomotion [51, 52], while simultaneously being capable of transmitting forces up to eight times the body weight during running [56]. Its function reduces the metabolic cost of overground movement and enhances propulsion [2, 54]. Achilles tendon rupture disrupts these functions, while typically causing chronic tendon elongation, more or less regardless of surgical or non-surgical treatment [140], accompanied by reduced stiffness [141, 142], and persistent plantar flexor weakness [50, 143, 144].

One year after an Achilles tendon rupture, walking was characterized by increased gastrocnemius activation and reduced ankle sagittal ROM in the affected shank compared with the contralateral side. During running, ankle ROM was likewise reduced and accompanied by attenuated plantar flexor moments, although no clear inter-limb differences in terms of EMG were observed. Despite these shank-level asymmetries, the overall gait patterns resembled normative kinematics, probably reflecting compensatory strategies [145]. EMG amplitudes and joint moments displayed greater variability than kinematic measures. These findings confirm persistent deficits in triceps surae function and tendon stiffness, consistent with previous reports [73, 78, 146]. Compensation strategies were found to vary, with some participants increasing knee extensor moments to shift propulsion proximally, while others maintained symmetrical knee kinetics, despite ankle impairments as discussed in detail above.

Integration Gap – Distribution of adaptations from feet up to trunk (Study IV)

If the biomechanical chain operated as a rigidly linked system, distal impairments like ATR would be expected to cause systematic proximal adaptations. However, the extent of such propagation depends on both mechanical attenuation-loss of force transmission with distance-and central nervous system strategies, which may buffer local disturbances.

In the present investigation, ATR-related adaptations were predominantly localized to the distal limb and expressed as side-to-side differences from the foot upward. The affected limb showed persistent ankle deficits and occasional compensations at the knee, whereas no consistent asymmetries were observed at the hip, pelvis, or trunk. One year after an Achilles tendon rupture, individuals thus presented task- and phase-specific neuromechanical asymmetries during walking, primarily at the ankle. This might reflect altered tendon mechanics and disrupted activation patterns, with a potential shift from monoarticular to biarticular muscle use, reduced activation variability, and redistribution of joint power across the lower limb. Such pattern supports a segmentally constrained model in which coupling is strongest between the adjacent segments and diminishes with distance from the site of impairment. Contralateral contributions and preserved proximal function may reduce the need for widespread reorganization.

RETHINKING THE BIOMECHANICAL CHAIN: ABSORPTIVE COUPLING MODEL

The integrated findings from Studies I–IV support an alternative view to the traditional Biomechanical Chain Model. In this *Absorptive Coupling Model*, mechanical effects are most pronounced at the site of change and in adjacent segments, while they progressively diminish at more proximally levels (Figure 51). Neural control mechanisms act in parallel to preserve overall coordination, limiting the spread of distal alterations throughout the chain. Distal changes can influence proximal function, yet the magnitude, direction, and consequences depend on task demands, individual variability, and the nervous system's compensatory capacity. The biomechanical chain is therefore neither a rigid sequence nor a purely local phenomenon, but a dynamic, probabilistic interaction system in which mechanical influences are absorbed along the way and coordination emerges from the interplay of mechanical and neural factors.

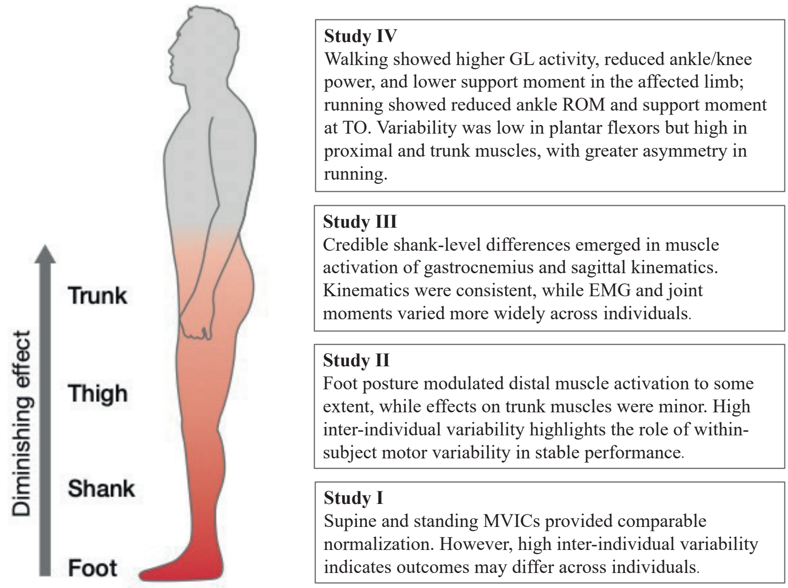

FIGURE 51. The absorptive coupling model of the biomechanical chain. In this model, local and adjacent segments exhibit the strongest mechanical coupling, while proximal influence progressively diminishes, with neural control maintaining overall coordination. GL = lateral gastrocnemius, TO = toe-off, EMG = electromyography, MVIC =maximum voluntary isometric contraction.

Three principles underpin the absorptive coupling model:

- *Neuromuscular variability is intrinsic*: activation patterns varied markedly between individuals across all tasks, consistent with motor abundance. This variability explains why some participants adopted knee-dominant propulsion after ATR, while others did not, and why foot posture changes had inconsistent trunk effects.
- *Adaptations are segment-specific*: the largest effects occurred locally and in adjacent segments, with clear attenuation along the chain.
- *Task and phase specificity matters*: distal–proximal effects were more pronounced in walking than running, likely due to stance-phase duration, reliance on push-off, and Achilles tendon elastic energy storage and return.

From a mechanical perspective, distal changes-whether experimentally induced (Study II) or related to ATR (Studies III–IV) produced the most consistent effects both locally and in neighboring segments. The impact generated at foot strike is progressively attenuated with increasing distance from its point of origin, likely due to joint compliance, soft tissue deformation, and energy dissipation within multi-joint muscle–tendon units. From a neural control perspective, the CNS uses motor abundance to achieve the same outcome through multiple coordination strategies, allowing local adjustments to preserve proximal stability and limiting widespread reorganization.

This integrated interpretation explains why trunk adaptations were minimal in both experimental and clinical contexts, and why compensations were often managed locally or through contralateral limb contributions. The *Absorptive Coupling Model* can be considered a possible reframing of the biomechanical chain, emphasizing its selective and context-dependent nature rather than a deterministic cascade. Such an interpretation appears consistent with contemporary motor control theory, with regards to variability as a functional property supporting adaptability in complex and unpredictable environments [25, 147, 148].

METHODOLOGICAL CONSIDERATIONS

LIMITATIONS, VALIDITY, AND RELIABILITY OF EMG AND MOTION CAPTURE SYSTEM (MOCAP)

Surface EMG measures the spatial summation of MUAPs at the muscle surface [149]. Because these potentials must travel through layers of tissue before reaching the electrodes, the recorded signal becomes attenuated [10]. Electrode placement is therefore critical and should be performed by one or only a few trained investigators strictly adhering to standardized protocols as was done in the present study [92]. Individual anatomical variations, however, means that few people exactly match the configurations illustrated in anatomical atlases. In addition, movement of electrode wires, skin motion relative to the underlying muscle, and crosstalk from adjacent muscles can all affect the signal amplitude [149]. The presence of electrodes, wires, sensors, and tape may also alter a participant's natural motor pattern during task performance.

The interpretation of sEMG should therefore be guided by careful adherence to protocol and further an awareness of its limitations [91]. Despite these constraints, several studies support its value in terms of biomechanical research [124–126]. For example, a study on paraspinal muscle fatigue concluded that there is convincing evidence for sEMG's utility for such assessments [150]. Mitchell et al. reported intra-session reliability for trunk muscle sEMG, with ICCs ranging from 0.80 to 0.90 in able-bodied controls; in individuals with spinal cord injury, [151] and Bogey et al. found surface electrodes to be as repeatable as fine-wire electrodes during gait analysis for the soleus muscle [152]. Taken together, these findings indicate that when applied using standardized protocols and interpreted cautiously, sEMG can yield reliable, meaningful insights into neuromuscular function, particularly for evaluating muscle activation patterns and fatigue.

Similarly, the reliability of MOCAP has been assessed extensively. A primary limitation arises from soft tissue artefacts caused by muscle contraction, gravity, skin deformation, or sliding of markers attached to the skin [153]. Concurrent validity is highest in the sagittal plane (correlation coefficient >0.7), lower in the coronal plane (0.5-0.6), and lowest in the transverse plane (≤0.4), [154]. Validation studies often compare MOCAP with radiostereometric analysis (RSA), in which tantalum markers are implanted into cortical bone and tracked by stereoradiographs [154, 155]. The strongest correlations between these measurement modalities have been reported for hip flexion and abduction, while rotational measures show poorer agreement [155].

Fändriks et al. compared three marker sets against RSA for knee motion analysis [156]. Although sagittal-plane flexion–extension measures were consistent across all sets, each underestimated true skeletal motion. Frontal and transverse plane measures were highly inconsistent, making them unreliable for precise motion assessment [156]. These results emphasize that while sagittal-plane measures, especially flexion/extension, are reasonably robust, frontal and transverse plane kinematics are more susceptible to soft tissue artefacts and marker displacement and should therefore be interpreted with caution.

Across all acquisition systems in the gait laboratory, high-quality results depend on a sound understanding of musculoskeletal anatomy, strict adherence to protocol, and meticulous handling of instrumentation. Transparent reporting of methods and data-processing algorithms is therefore essential for reproducibility and for meaningful interpretation [92], particularly when studying the biomechanical chain from the foot to trunk.

ETHICAL CONSIDERATIONS

This thesis adheres to the principles of the Declaration of Helsinki [157] and the European Code of Conduct for Research Integrity [158], upholding reliability, honesty, respect, and accountability throughout the entire research process.

The positive outcomes of all four studies were considered to outweigh any potential discomfort experienced by the study participants. Given that the methodologies employed were non-invasive, the burden placed on participants can be considered minimal.

Biomechanical data collection often requires participants to wear less clothing (e.g., shorts and a top), which may be perceived as somewhat uncomfortable or awkward by some individuals. The data collection process can be time-consuming, as sEMG electrode placement requires thorough skin preparation, which may occasionally cause minor skin irritation. No participants reported complaints, and no dropouts occurred due to personal issues related to the method during the laboratory investigations. Also, the isometric contractions performed during the sEMG session can cause some muscle soreness during one or two days after the testing session. All participants were fully informed of these procedures beforehand, and informed consent was always obtained. The studies involved sensitive personal data related to health and well-being; however, all data were pseudo-anonymized, and the coding key was securely stored and accessible only to the responsible researcher.

STATISTICAL CONSIDERATIONS

The four studies applied statistical methods selected not only for their suitability to each aim but also for their capacity to probe the biomechanical chain from multiple angles. Studies I–II used frequentist tests for controlled EMG comparisons; Study III employed Bayesian hierarchical modelling to capture limb-to-limb asymmetries and quantify uncertainty in gait after an Achilles tendon rupture; and Study IV applied statistical parametric mapping to reveal time-specific differences in continuous kinetic and EMG profiles.

This methodological diversity reflects the exploratory incentive of the thesis; a deliberate effort to apply statistical tools capable of uncovering both broad group-level patterns and more subtle, individualized adaptations. By integrating frequentist, Bayesian, and time-series approaches, the work achieved a multifaceted perspective on neuromechanical function, enabling a richer and more precise exploration of how mechanical and neural influences interact along the biomechanical chain.

GENDER CONSIDERATION

Participant sex was recorded; however, the limited number of participants precluded meaningful subgroup analyses, which were also beyond the scope of the present studies. Although sex-related aspects of Achilles tendon rupture have been explored [159-161], they warrant more focused attention in future research.

9. Strengths and limitations

"

Every morning in Africa, a gazelle wakes up, it knows it must outrun the fastest lion or it will be killed. Every morning in Africa, a lion wakes up. It knows it must run faster than the slowest gazelle, or it will starve. It doesn't matter whether you're the lion or a gazelle-when the sun comes up, you'd better be running.

CHRISTOPHER MCDOUGALL

STRENGTHS

A major strength of this thesis lies in its comprehensive investigation of the biomechanical chain from the foot to the trunk, integrating EMG, kinematic, and kinetic data across multiple functional tasks. This multilevel approach offers novel insight into both methodological and clinical aspects of segmental coordination during human movement.

First, the inclusion of two methodological studies (Studies I and II) served to strengthen the overall methodological framework by addressing critical aspects of EMG normalization and foot positioning. The standardized test protocols, including consistent MVIC procedures and electrode placements based on established anatomical landmarks, enhanced the reliability and reproducibility of the EMG data used in the subsequent ATR-related investigations.

Second, the application of ecologically valid motor tasks, such as stair climbing, walking, and jogging, increases the translational relevance of the findings. The inclusion of different foot positions and gait speeds allowed for a nuanced exploration of how neuromechanical strategies may vary under changing mechanical demands, meaning variations in external loading and movement constraints imposed by speed and foot alignment.

Third, the use of advanced motion capture technologies, including a high-resolution optical tracking system synchronized with force plates, enabled precise analysis of joint kinetics and support moments using inverse dynamics. These biomechanical tools allowed for time-continuous, segment-specific analyses that are rarely integrated in clinical gait research, particularly in relation to ATR.

Finally, Studies III and IV represent an effort to explore locomotor adaptations after an ATR, beyond the ankle, providing detailed descriptions of interlimb differences in muscle activation, ROM and support strategies during walking and running. The integration of

EMG, joint kinetics, and kinematics support a deep understanding of segmental compensation patterns in this clinical population.

LIMITATIONS

Despite its strengths, this work has several limitations. The sample sizes, typical for the gait EMG studies, were large enough for within-subject analyses but limit generalizability, particularly in terms of inter-individual variability. In Study IV, the inclusion of both surgically and non-surgically treated ATR participants introduced clinical heterogeneity that was not stratified in the analysis.

The cross-sectional design of Studies III and IV precludes conclusions about the temporal progression of compensatory strategies following ATR. Longitudinal data are needed to determine whether observed adaptations persist, resolve, or evolve over time.

Laboratory-based tasks, while controlled, may not fully reflect sport-specific or real-world movement demands. Foot posture manipulations in Study II, for example, although allowing individual adaptation, may not completely replicate natural pronation or supination of the foot.

Methodologically, surface EMG, though widely accepted, remains vulnerable to signal attenuation, cross-talk, and skin movement artifacts, particularly for deep or small muscles. Likewise, skin-mounted optical markers, despite high spatial resolution, can be affected by soft tissue artifacts, especially in the frontal and transverse planes, which may reduce the precision of joint angle estimates.

Although task execution was standardized with instructions and practice trials, individual movement strategies may still have influenced outcomes. Awareness of instrumentation (e.g., electrodes, markers) may also – at least slightly – have altered performance.

Psychological factors such as fear of reinjury or reduced confidence were discussed conceptually but not measured, representing an important avenue for future research with a biopsychosocial approach.

Finally, biomechanical tools quantify parameters such as muscle activation, joint angles, and forces with high precision, but offer only a partial view of functional capacity. Laboratory measurements cannot fully capture the complexity, variability, and adaptability of movements in natural environments. Thus, it is suggested that biomechanical analysis should be interpreted as part of a broader framework integrating ecological, psychological, and functional perspectives.

10. Conclusion

"

One of the wonderful aspects of running is that there is no definition of a 'runner' that you must live up to.

KARA GOUCHER

From the integrated findings of the present thesis, several overarching insights emerge. The obtained data contribute to improve our understanding of how the biomechanical chain operates in both healthy and post-injury contexts, and they may provide a potential framework for both future research strategies and possibly improved understanding of given rehabilitation strategies:

- Distal changes dominate but may propagate proximally

Neuromechanical effects are strongest locally and in adjacent segments, with attenuation observed more proximally; their impact depends on task demands, individual variability, and compensatory capacity.

- Neuromuscular variability is intrinsic and functional

High within- and between-individual variability reflects motor abundance, allowing exploration of alternative coordination strategies to maintain performance despite constraints.

- Adaptations are context- and phase-specific

Distal–to-proximal effects are more evident in walking than running, shaped by stance duration, push-off demands, and the elastic energy use of the Achilles tendon.

- Adaptive compensations after an ATR remain localized

Even one year post-injury, kinetic and EMG asymmetries are most apparent distally, with no consistent changes at the hip or trunk, suggesting that rehabilitation may be particularly relevant at the local level.

- The biomechanical chain is dynamic, not rigid

11. Clinical implication

"

Han kunde tydligt se Snusmumrikens fotspår i den våta jorden. De fnattade hit och dit och var ganska svåra att följa. Ibland gjorde de långa skutt och korsade sig själva. Han har varit glad, funderade Mumintrollet. Här har han gjort en kullerbytta, det är klart och tydligt.
He could see Snufkin's footprints clearly in the damp earth. They zigzagged this way and that, tricky to follow. Now and then they leapt far, sometimes even crossing themselves. He's been in high spirits, thought Moomintroll. Here he's turned a somersault, no doubt about it.

TOVE JANSSON

It is well established that rehabilitation is grounded in clinical reasoning, emphasizing individualized, task-specific loading, functional progression, and continuous outcome assessment guided by the specific and individualized patient goals [161-165].The clinical implications of this thesis build on this foundation and further, the aim to promote an even more individualized rehabilitation approach and potentially providing new insights for practice.

As with all tendon repair, healing is a slow process, and this needs to be carefully respected when building a suitable and resilient tendon [166]. In harmony with previously suggested rehabilitation progression in terms of an ATR [62, 72], post-injury rehabilitation is likely to benefit from being even more individually tailored, progressively loaded, and finally task-specific. The focus might therefore not only be related to restoring plantar flexor strength, but also on enhancing neuromuscular coordination, tendon stiffness, and resilience across the entire kinetic chain.

Guided by the proposed *Absorptive Coupling Model*, this process may recognize that mechanical effects of an Achilles tendon rupture are strongest locally and in adjacent segments, while they attenuate more proximally, with neural control maintaining overall coordination. Consequently, and in harmony with this model, the rehabilitation program would therefore need to address local tendon capacity and adjacent segment reconditioning, while ensuring effective reintegration of proximal structures for an efficient whole-chain function.

While structured programs provide valuable guidance, rehabilitation may also be enhanced by viewing it as a dynamic capacity-building process, with attentiveness to when individual deviations from the structure may become necessary. Loading can be systematically and gradually progressed with the goal to stimulate collagen remodeling, optimize stiffness, and refine intersegmental coordination. This progression might then be individually calibrated according to the tendon healing stage, structural integrity, neuromuscular control, and psychological readiness. Variability in load magnitude, direction, and task-specific context will challenge both mechanical and neural systems, enhancing adaptability and resilience to the often unpredictable real-world demands.

Based on the findings of the present thesis, together with previously established principles, the following are suggested as potential guiding principles.

- *Individually-tailored rehabilitation protocols*: adjust the protocol-by taking into account the biological healing process to each patient's biomechanics, goals, and recovery rate.
- *Progressive load:* gradually increase the demands to promote tendon adaptation and structural resilience.
- *Task-specific interventions*: Use functional, context-rich activities for daily life and sports activity.
- *Integrated kinetic chain focus*: target plantar flexor strength, neuromuscular coordination, and multi-segment reintegration.
- *Normalized movement to the individual: g*uide toward efficient and sustainable strategies while respecting individual variability.
- *Shape motor variability*: encourage adaptive flexibility, avoid maladaptive patterns.
- *Continuously adjustable*: modify the response according to the evolving structural and coordination capacity.

In terms of this integrated approach, successful rehabilitation may not be defined solely by restoring the symmetry or baseline gait, but by developing a resilient and dynamic tendon–muscle–neural system capable of sustaining high performance and adapting to the individually varying demands of sport, work, and daily life.

12. Future perspectives

"

Most runners hope running will always be a part of their lives. I'll be happy if running and I can grow old together.

HARUKI MURAKAMI

The findings of this thesis highlight several avenues for future investigation spanning methodological, experimental, and applied domains. Addressing these domains will deepen the understanding of the biomechanical chain and improve clinical outcomes for individuals with distal impairments such as an ATR.

LONGITUDINAL TRACKING OF RECOVERY TRAJECTORIES

Future studies are suggested to be able to follow ATR recovery from the acute phase through long-term outcomes, mapping changes in joint kinetics, kinematics, and EMG patterns. Such tracking would help identify critical intervention windows and determine whether observed compensations resolve, persist, or develop into maladaptive strategies that affect efficiency, load distribution, or secondary joint health.

INTEGRATION OF PSYCHOLOGICAL FACTORS WITH BIOMECHANICS

Fear of re-injury, reduced limb confidence, and perceived instability may likely shape post-ATR movement strategies. Combining validated psychological assessments (e.g., kinesiophobia scales) with biomechanical, EMG measures and functional scores could improve predictive models of adaptation. A biopsychosocial approach would support interventions that address both mechanical deficits and movement confidence.

DYNAMIC EVALUATION OF ORTHOSES AND FOOTWEAR

Many orthoses and footwear products claim to influence the kinetic chain beyond the foot, yet evidence under dynamic, real-world conditions is scarce. Building on the present thesis, future research efforts should implement and evaluate these intervention strategies in walking, running, and sport-specific tasks, quantifying local and proximal adaptations and their impact on propulsion, stability, and load management.

PERTURBATION-BASED ASSESSMENTS OF CHAIN ROBUSTNESS

Steady-state gait may mask distal–proximal coupling. Introducing controlled perturbations such as surface changes, unexpected loads, or directional shifts could reveal latent mechanical and neural linkages. Such testing could identify individuals with reduced adaptability, who may be at higher risk of reinjury.

DEVELOPMENT OF UPDATED NORMATIVE DATASETS

Current gait and EMG reference datasets are often outdated or based on small, homogeneous samples. Thus, there is a need for establishing diverse normative datasets that include joint power, muscle activation, and variability metrics across walking, running, and other functional tasks to improve benchmarking and interpretation for post-injury assessments.

13. Acknowledgement

"

Nog finns det mål och mening i vår färd—
men det är vägen, som är mödan värd.

Indeed, there may be aim and goal to our stride—
but it is the road that makes the effort abide.

KARIN BOYE

This thesis has been a shared adventure. Many people contributed along the way. First and foremost, a heartfelt thank you to all the **study participants**. Your endless patience was exceptional. Your participation made this work possible.

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* T.S. Eliot

** Robert Frost

*** Morticia Addams, by Charles Addams

**** 1 Corinthians 13:13

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14. References

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Say not, 'I have found the truth,' but rather, 'I have found a truth.' Say not, 'I have found the path of the soul.' Say rather, 'I have met the soul walking upon my path.

KAHLIL GIBRAN

1. Winter DA. Biomechanics and Motor Control of Human Movement. 4th ed. New Jersey: John Wiely and Sons; 2009.

2. Alexander RM. Energy-saving mechanisms in walking and running. Journal of experimental biology. 1991;160(1):55-69.

3. Gardner GE. The Story of the Human Body: Evolution, Health, and Disease, by Daniel E. Lieberman. University of California Press USA; 2014.

4. Shumway-Cook A, Woollacott MH. Motor Control, Translanting Research into Clinical Practice. 3rd ed. Baltimore: Lippincott Williams and Wilikins; 2007.

5. Gray H, Carter HV, Ukray M. Gray's Anatomy:(Illustrated With 1247 Coloured Well Drawing Engrawings). E-Kitap Projesi & Cheapest Books; 2023.

6. Latash ML. Fundamentals of motor control. Academic Press; 2012.

7. World Confederation for Physical Therapy. Policy statement: description of physical therapy. World Confederation for Physical Therapy, 2011. https://world.physio/policy/ps-descriptionPT

8. APTA. Standards of Practice for Physical Therapy. 2020. https://www.apta.org/siteassets/pdfs/policies/standards-of-practice-pt.pdf

9. World Confederation for Physical Therapy. "Description of physical therapy: Policy statement." (2019). https://world.physio/sites/default/files/2020-04/PS-2019-Description-of-physical-therapy.pdf

10. Basmajian J, de Luca CJ. Muscles alive. Their Function Revealed by Electromyography. 5th ed. Baltimore, USA: Williams and Wilkins; 1985.

11. Everett T, Kell C. Human movement: an introductory text. 6th ed. Edinburgh: Churchill Livingstone; 2010.

12. Piscitelli D, Falaki A, Solnik S, Latash ML. Anticipatory postural adjustments and anticipatory synergy adjustments: preparing to a postural perturbation with predictable and unpredictable direction. Exp Brain Res. 2017;235(3):713-30.

13. Assaiante C, Mallau S, Viel S, Jover M, Schmitz C. Development of postural control in healthy children: a functional approach. Neural Plast. 2005;12(2-3):109-18; discussion 263-72.

14. Müller H, Sternad D. Motor learning: changes in the structure of variability in a redundant task. Adv Exp Med Biol. 2009;629:439-56.

15. Fetters L. Perspective on variability in the development of human action. Phys Ther. 2010;90(12):1860-7.

16. Madeleine P, Voigt M, Mathiassen SE. The size of cycle-to-cycle variability in biomechanical exposure among butchers performing a standardised cutting task. Ergonomics. 2008;51(7):1078-95.

17. Komar J, Seifert L, Thouvarecq R. What variability tells us about motor expertise: measurements and perspectives from a complex system approach. Mov Sport Sci Sci Motric. 2015;89(3):65-77.

18. Srinivasan D, Rudolfsson T, Mathiassen SE. Between- and within-subject variance of motor variability metrics in females performing repetitive upper-extremity precision work. J Electromyogr Kinesiol. 2015;25(1):121-9.

19. Hatze H. Motion variability-its definition, quantification, and origin. J Mot Behav. 1986;18(1):5-16.

20. Latash ML. The bliss (not the problem) of motor abundance (not redundancy). Experimental brain research. 2012;217(1):1-5.

21. Bongaardt R, Meijer OG. Bernstein's theory of movement behavior: historical development and contemporary relevance. J Mot Behav. 2000;32(1):57-71.

22. Davids K, Bennett S, Newell KM. Movement system variability. Champaign (IL): Human Kinetics; 2006.

23. Riley MA, Turvey MT. Variability and determinism in motor behavior. J Mot Behav. 2002;34(2):99-125.

24. Bartlett R, Wheat J, Robins M. Is movement variability important for sports biomechanists? Sports Biomech. 2007;6(2):224-43.

25. Simonsen EB, Alkjær T. The variability problem of normal human walking. Med Eng Phys. 2012;34(2):219-24.

26. Abboud J, Nougarou F, Pagé I, Cantin V, Massicotte D, Descarreaux M. Trunk motor variability in patients with non-specific chronic low back pain. Eur J Appl Physiol. 2014;114(12):2645-54.

27. Madeleine P, Farina D. Time to task failure in shoulder elevation is associated to increase in amplitude and to spatial heterogeneity of upper trapezius mechanomyographic signals. Eur J Appl Physiol. 2008;102(3):325-33.

28. Farina D, Merletti R, Enoka RM. The extraction of neural strategies from the surface EMG. J Appl Physiol (1985). 2004;96(4):1486-95.

29. Jandacka D, Blaschova D, Amado A, van Emmerik R, Silvernail JF, Hamill J. Coordination variability in runners after surgical Achilles tendon repair. Translational Sports Medicine. 2021;4(2):204-13.

30. Levangie PK, Norkin CC. Joint structure and function: a comprehensive analysis. 5th ed. Philadelphia: F.A. Davis; 2011.

31. Smith LK, Weiss EL, Lehmkuhl LD. Brunnstrom's clinical kinesiology. 5th ed. Philadelphia: F.A. Davis Company; 1996.

32. Perry J, Burnfield J. Gait Analysis: Normal and Pathological Function. 2nd ed. Boca Raton: CRC Press; 2024.

33. Saeterbakken AH, Fimland MS. Muscle activity of the core during bilateral, unilateral, seated and standing resistance exercise. European journal of applied physiology. 2012;112(5):1671-8.

34. Escamilla RF, Lewis C, Pecson A, Imamura R, Andrews JR. Muscle activation among supine, prone, and side position exercises with and without a Swiss ball. Sports Health. 2016;8(4):372-9.

35. Clarke B, Al-Hammdany JK, Di Giulio I. Human muscle and spinal activation in response to body weight loading. J Anat. 2023;242(5):745-53.

36. Kapandji A, Owerko C, Anderson A. The physiology of the joints. 7th ed. London: Handspring Publishing Limited; 2019.

37. Alexander RM. Simple models of human movement. London: Taylor & Francis; 1995.

38. Kuo AD, Donelan JM, Ruina A. Energetic consequences of walking like an inverted pendulum: step-to-step transitions. Exercise and sport sciences reviews. 2005;33(2):88-97.

39. Novacheck TF. The biomechanics of running. Gait Posture. 1998;7(1):77-95.

40. Cavagna GA, Saibene FP, Margaria R. Mechanical work in running. Journal of applied physiology. 1964;19(2):249-56.

41. Hoitz F, von Tscharner V, Baltich J, Nigg BM. Individuality decoded by running patterns: movement characteristics that determine the uniqueness of human running. PLoS One. 2021;16(4):e0249657.

42. Bramble DM, Lieberman DE. Endurance running and the evolution of Homo. Nature. 2004;432(7015):345-52.

43. Ker R, Bennett M, Bibby S, Kester R, Alexander RM. The spring in the arch of the human foot. Nature. 1987;325(6100):147-9.

44. Nigg BM, Herzog W. Biomechanics of the musculoskeletal system. 3rd ed. Chichester (UK): John Wiley & Sons Ltd; 2006.

45. Standring S, Ellis H, Healy JC, Johnson D, Williams A, Collins P, et al. Gray's anatomy: the anatomical basis of clinical practice. 39th ed. Edinburgh: Elsevier Churchill Livingstone; 2005.

46. Levine D, Whittle MW. The effects of pelvic movement on lumbar lordosis in the standing position. J Orthop Sports Phys Ther. 1996;24(3):130-5.

47. Yoo HJ, Sim T, Choi A, Park HJ, Yang H, Heo HM, et al. Quantifying coordination between agonist and antagonist muscles during gait. J Mech Sci Technol. 2016;30(11):5321-8.

48. Tengman T, Riad J. Three-dimensional gait analysis following Achilles tendon rupture with nonsurgical treatment reveals long-term deficiencies in muscle strength and function. Orthop J Sports Med. 2013;1(4):2325967113504734.

49. Willy RW, Brorsson A, Powell HC, Willson JD, Tranberg R, Grävare Silbernagel K. Elevated knee joint kinetics and reduced ankle kinetics are present during jogging and hopping after Achilles tendon ruptures. Am J Sports Med. 2017;45(5):1124-33.

50. Don R, Ranavolo A, Cacchio A, Serrao M, Costabile F, Iachelli M, et al. Relationship between recovery of calf muscle biomechanical properties and gait pattern following surgery for Achilles tendon rupture. Clin Biomech. 2007;22(2):211-20.

51. Bojsen-Møller J, Magnusson SP. Heterogeneous loading of the human Achilles tendon in vivo. Exerc Sport Sci Rev. 2015;43(4):190-7.

52. Józsa L, Kannus P. Human tendons: anatomy, physiology, and pathology. Champaign (IL): Human Kinetics; 1997.

53. Malvankar S, Khan WS. Evolution of the Achilles tendon: the athlete's Achilles heel? Foot (Edinb). 2011;21(4):193-7.

54. Fukunaga T, Kubo K, Kawakami Y, Fukashiro S, Kanehisa H, Maganaris CN. In vivo behaviour of human muscle tendon during walking. Proc Biol Sci. 2001;268(1464):229-33.

55. Ishikawa M, Komi PV, Grey MJ, Lepola V, Brüggemann GP. Muscle–tendon interaction and elastic energy usage in human walking. J Appl Physiol (1985). 2005;99(2):603-8.

56. Komi PV, Fukashiro S, Järvinen M. Biomechanical loading of Achilles tendon during normal locomotion. Clin Sports Med. 1992;11(3):521-31.

57. Maffulli N, Almekinders LC. The Achilles tendon. London: Springer; 2007.

58. Edama M, Kubo M, Onishi H, Takabayashi T, Inai T, Yokoyama E, et al. The twisted structure of the human Achilles tendon. Scand J Med Sci Sports. 2015;25(5):497-503.

59. Pękala P, Henry B, Ochała A, Kopacz P, Tatoń G, Młyniec A, et al. The twisted structure of the Achilles tendon unraveled: a detailed quantitative and qualitative anatomical investigation. Scand J Med Sci Sports. 2017;27(12):1705-15.

60. Arndt A, Bengtsson AS, Peolsson M, Thorstensson A, Movin T. Non-uniform displacement within the Achilles tendon during passive ankle joint motion. Knee Surg Sports Traumatol Arthrosc. 2012;20(9):1868-74.

61. Magnusson SP, Narici MV, Maganaris CN, Kjaer M. Human tendon behaviour and adaptation, in vivo. J Physiol. 2008;586(1):71-81.

62. Tarantino D, Palermi S, Sirico F, Corrado B. Achilles tendon rupture: mechanisms of injury, principles of rehabilitation and return to play. J Funct Morphol Kinesiol. 2020;5(4):99.

63. Campos PT, da Costa MQ, Lascano GCM. Achilles tendon rupture: discussion and updates [Ruptura do tendão de Aquiles: discussão e atualizações]. J Foot Ankle. 2024;18(3):308-14.

64. Kannus P, Józsa L. Histopathological changes preceding spontaneous rupture of a tendon: a controlled study of 891 patients. J Bone Joint Surg Am. 1991;73(10):1507-25.

65. Ahmed I, Lagopoulos M, McConnell P, Soames R, Sefton G. Blood supply of the Achilles tendon. J Orthop Res. 1998;16(5):591-6.

66. Ganestam A, Kallemose T, Troelsen A, Barfod KW. Increasing incidence of acute Achilles tendon rupture and a noticeable decline in surgical treatment from 1994 to 2013: a nationwide registry study of 33,160 patients. Knee Surg Sports Traumatol Arthrosc. 2016;24(12):3730-7.

67. Svedman S, Marcano A, Ackermann PW, Felländer-Tsai L, Berg HE. Acute Achilles tendon ruptures between 2002 and 2021: sustained increased incidence, surgical decline and prolonged delay to surgery: a nationwide study of 53,688 ruptures in Sweden. BMJ Open Sport Exerc Med. 2024;10(3):e001960.

68. Leino O, Keskinen H, Laaksonen I, Mäkelä K, Löyttyniemi E, Ekman E. Incidence and treatment trends of Achilles tendon ruptures in Finland: a nationwide study. Orthop J Sports Med. 2022;10(11):23259671221131536.

69. Briggs-Price S, Mangwani J, Houchen-Wolloff L, Modha G, Fitzpatrick E, Faizi M, et al. Incidence, demographics, characteristics and management of acute Achilles tendon rupture: an epidemiological study. PLoS One. 2024;19(6):e0304197.

70. Vosseller JT, Ellis SJ, Levine DS, Kennedy JG, Elliott AJ, Deland JT, et al. Achilles tendon rupture in women. Foot Ankle Int. 2013;34(1):49-53.

71. Hartman H, Cacace A, Leatherman H, Ashkani-Esfahani S, Guss D, Waryasz G, et al. Gender differences in Achilles tendon ruptures: a retrospective study and a review of the literature. J Foot Ankle Surg. 2024;63(5):614-20.

72. Brorsson A. Acute Achilles tendon rupture: the impact of calf muscle performance on function and recovery [dissertation]. Gothenburg (Sweden): University of Gothenburg; 2017.

73. Brorsson A, Grävare Silbernagel K, Olsson N, Nilsson Helander K. Calf muscle performance deficits remain 7 years after an Achilles tendon rupture. Am J Sports Med. 2018;46(2):470-7.

74. Brorsson A, Willy RW, Tranberg R, Grävare Silbernagel K. Heel-rise height deficit 1 year after Achilles tendon rupture relates to changes in ankle biomechanics 6 years after injury. Am J Sports Med. 2017;45(13):3060-8.

75. Silbernagel KG, Steele R, Manal K. Deficits in heel-rise height and Achilles tendon elongation occur in patients recovering from an Achilles tendon rupture. Am J Sports Med. 2012;40(7):1564-71.

76. Kangas J, Pajala A, Ohtonen P, Leppilahti J. Achilles tendon elongation after rupture repair: a randomized comparison of two postoperative regimens. Am J Sports Med. 2007;35(1):59-64.

77. Zhang LN, Wan WB, Wang YX, Jiao ZY, Zhang LH, Luo YK, et al. Evaluation of elastic stiffness in healing Achilles tendon after surgical repair of a tendon rupture using in vivo ultrasound shear wave elastography. Med Sci Monit. 2016;22:1186-91.

78. Khair RM, Watt J, Subanen M, Cronin NJ, Finni T. Neuromechanical adaptations in the gastrocnemius muscle after Achilles tendon rupture during walking. J Electromyogr Kinesiol. 2025;80:102962.

79. Zellers JA, Marmon AR, Ebrahimi A, Grävare Silbernagel K. Lower extremity work along with triceps surae structure and activation is altered with jumping after Achilles tendon repair. J Orthop Res. 2019;37(4):933-41.

80. Oda H, Sano K, Kunimasa Y, Komi PV, Ishikawa M. Neuromechanical modulation of the Achilles tendon during bilateral hopping in patients with unilateral Achilles tendon rupture over 1 year after surgical repair. Sports Med. 2017;47(6):1221-30.

81. Jandacka D, Plesek J, Skypala J, Uchytil J, Silvernail JF, Hamill J. Knee joint kinematics and kinetics during walking and running after surgical Achilles tendon repair. Orthop J Sports Med. 2018;6(6):2325967118779862.

82. Jandacka D, Silvernail JF, Uchytil J, Zahradnik D, Farana R, Hamill J. Do athletes alter their running mechanics after an Achilles tendon rupture? J Foot Ankle Res. 2017;10:53.

83. Elkjær E, Mikkelsen MB, Michalak J, Mennin DS, O'Toole MS. Expansive and contractive postures and movement: a systematic review and meta-analysis of the effect of motor displays on affective and behavioral responses. Perspect Psychol Sci. 2022;17(1):276-304.

84. Larsson E, LeGreves A, Brorsson A, Eliasson P, Johansson C, Carmont MR, et al. Fear of reinjury after acute Achilles tendon rupture is related to poorer recovery and lower physical activity postinjury. J Exp Orthop. 2024;11(4):70.

85. Jónsdóttir US, Briem K, Tranberg R, Brorsson A. The effect of fear of reinjury on joint power distribution during a drop countermovement jump two years after an Achilles tendon rupture. Transl Sports Med. 2021;4(5):667-74.

86. Ayers DC, Franklin PD, Ring DC. The role of emotional health in functional outcomes after orthopaedic surgery: extending the biopsychosocial model to orthopaedics: AOA critical issues. J Bone Joint Surg Am. 2013;95(21):e165.

87. Jónsdóttir US, Brorsson A, Nilsson Helander K, Tranberg R, Larsson ME. Factors that affect return to sports after an Achilles tendon rupture: a qualitative content analysis. Orthop J Sports Med. 2023;11(2):23259671221145199.

88. Kamen G, Caldwell GE. Physiology and interpretation of the electromyogram. J Clin Neurophysiol. 1996;13(5):366-84.

89. Merletti R, Botter A, Troiano A, Merlo E, Minetto MA. Technology and instrumentation for detection and conditioning of the surface electromyographic signal: state of the art. Clin Biomech. 2009;24(2):122-34.

90. Burden A. Surface electromyography. In: Payton CJ, Bartlett RM, editors. Biomechanical evaluation of movement in sport and exercise. London: Routledge; 2007.

91. Criswell E. Cram's introduction to surface electromyography. 2nd ed. Sudbury (MA): Jones & Bartlett Publishers; 2010.

92. Hermens HJ, Freriks B, Disselhorst-Klug C, Rau G. Development of recommendations for SEMG sensors and sensor placement procedures. J Electromyogr Kinesiol. 2000;10(5):361-74.

93. Konrad P. The ABC of EMG: a practical introduction to kinesiological electromyography. Version 1.0. Noraxon USA Inc.; 2005.

94. Barbero M, Merletti R, Rainoldi A. Atlas of muscle innervation zones: understanding surface electromyography and its applications. Milan: Springer; 2012.

95. Karlsson JS, Roeleveld K, Grönlund C, Holtermann A, Östlund N. Signal processing of the surface electromyogram to gain insight into neuromuscular physiology. Philos Trans A Math Phys Eng Sci. 2009;367(1887):337-56.

96. De Luca CJ, Gilmore LD, Kuznetsov M, Roy SH. Filtering the surface EMG signal: movement artifact and baseline noise contamination. J Biomech. 2010;43(8):1573-9.

97. Farina D, Merletti R, Enoka RM. The extraction of neural strategies from the surface EMG: an update. J Appl Physiol (1985). 2014;117(11):1215-30.

98. Burden A. How should we normalize electromyograms obtained from healthy participants? What we have learned from over 25 years of research. J Electromyogr Kinesiol. 2010;20(6):1023-35.

99. Cappozzo A, Catani F, Leardini A, Benedetti M, Della Croce U. Position and orientation in space of bones during movement: experimental artefacts. Clin Biomech. 1996;11(2):90-100.

100. Kopellovich D. Shore (durometer) hardness test [Internet]. https://www.substech.com/dokuwiki/doku.php?id=shore_durometer_hardness_test

101. Brody DM. Techniques in the evaluation and treatment of the injured runner. Orthop Clin North Am. 1982;13(3):541-58.

102. Menz HB. Alternative techniques for the clinical assessment of foot pronation. J Am Podiatr Med Assoc. 1998;88(3):119-29.

103. McPoil T, Cornwall M, Abeler M, Devereaux K, Flood L, Merriman S, et al. The optimal method to assess the vertical mobility of the midfoot: navicular drop versus dorsal arch height difference. Clin Res Foot Ankle. 2013;1(1):1-7.

104. Winter DA. Overall principle of lower limb support during stance phase of gait. J Biomech. 1980;13(11):923-7.

105. Simonsen E, Dyhre-Poulsen P, Voigt M, Aagaard P, Fallentin N. Mechanisms contributing to different joint moments observed during human walking. Scand J Med Sci Sports. 1997;7(1):1-13.

106. Devlin NJ, Brooks R. EQ-5D and the EuroQol group: past, present and future. Appl Health Econ Health Policy. 2017;15(2):127-37.

107. Hung MC, Lu WS, Chen SS, Hou WH, Hsieh CL, Wang JD. Validation of the EQ-5D in patients with traumatic limb injury. J Occup Rehabil. 2015;25(2):387-93.

108. Saltin B, Grimby G. Physiological analysis of middle-aged and old former athletes: comparison with still active athletes of the same ages. Circulation. 1968;38(6):1104-15.

109. Kruschke JK, Liddell TM. Bayesian data analysis for newcomers. Psychon Bull Rev. 2018;25(1):155-77.

110. R Core Team. R: a language and environment for statistical computing [Internet]. Vienna (Austria): R Foundation for Statistical Computing; 2025 [cited 2025 Sept 16]. https://www.R-project.org/

111. Bürkner PC. brms: an R package for Bayesian multilevel models using Stan. J Stat Softw. 2017;80(1):1-28.

112. Vehtari A, Gabry J, Magnusson M, Yao Y, Bürkner P, Paananen T, et al. loo: efficient leave-one-out cross-validation and WAIC for Bayesian models. R package version 2.8.0. 2024. https://mc-stan.org/loo/

113. Hopkins WG. Measures of reliability in sports medicine and science. Sports Med. 2000;30(1):1-15.

114. Faul F, Erdfelder E, Lang AG, Buchner A. G*Power 3: a flexible statistical power analysis program for the social, behavioral, and biomedical sciences. Behav Res Methods. 2007;39(2):175-91.

115. Pataky TC. Generalized n-dimensional biomechanical field analysis using statistical parametric mapping. J Biomech. 2010;43(10):1976-82.

116. Penny WD, Friston KJ, Ashburner JT, Kiebel SJ, Nichols TE. Statistical parametric mapping: the analysis of functional brain images. London: Elsevier; 2011.

117. Bonett DG, Seier E. Confidence interval for a coefficient of dispersion in nonnormal distributions. Biom J. 2006;48(1):144-8.

118. Martens J, Daly D, Deschamps K, Staes F, Fernandes RJ. Inter-individual variability and pattern recognition of surface electromyography in front crawl swimming. J Electromyogr Kinesiol. 2016;31:14-21.

119. Souza TR, Pinto RZ, Trede RG, Kirkwood RN, Fonseca ST. Temporal couplings between rearfoot–shank complex and hip joint during walking. Clin Biomech. 2010;25(7):745-8.

120. Yazdani F, Razeghi M, Karimi MT, Raeisi Shahraki H, Salimi Bani M. The influence of foot hyperpronation on pelvic biomechanics during stance phase of gait: a biomechanical simulation study. Proc Inst Mech Eng H. 2018;232(7):708-17.

121. Khamis S, Yizhar Z. Effect of feet hyperpronation on pelvic alignment in a standing position. Gait Posture. 2007;25(1):127-34.

122. Ireland ML, Willson JD, Ballantyne BT, Davis IM. Hip strength in females with and without patellofemoral pain. J Orthop Sports Phys Ther. 2003;33(11):671-6.

123. Farahpour N, Jafarnezhadgero A, Allard P, Majlesi M. Muscle activity and kinetics of lower limbs during walking in pronated feet individuals with and without low back pain. J Electromyogr Kinesiol. 2018;39:35-41.

124. Jafarnezhadgero A, Ghane G, MokhtariMalekAbadi A, Valizadehorang A. Comparison of lower limb muscular activities during three different running patterns in pronated feet individuals with and without low back pain. Anesth Pain Med. 2020;11(4):e109724.

125. Alam MF, Ansari S, Zaki S, Sharma S, Nuhmani S, Alnagmoosh A, et al. Effects of physical interventions on pain and disability in chronic low back pain with pronated feet: a systematic review and meta-analysis. Physiother Theory Pract. 2025;41(2):390-404.

126. Pinto RZ, Souza TR, Trede RG, Kirkwood RN, Figueiredo EM, Fonseca ST. Bilateral and unilateral increases in calcaneal eversion affect pelvic alignment in standing position. Man Ther. 2008;13(6):513-9.

127. Betsch M, Schneppendahl J, Dor L, Jungbluth P, Grassmann JP, Windolf J, et al. Influence of foot positions on the spine and pelvis. Arthritis Care Res (Hoboken). 2011;63(12):1758-65.

128. Hornestam JF, Arantes PMM, Souza TR, Resende RA, Aquino CF, Fonseca ST, et al. Foot pronation affects pelvic motion during the loading response phase of gait. Braz J Phys Ther. 2021;25(6):727-34.

129. Gallagher KM, Wong A, Callaghan JP. Possible mechanisms for the reduction of low back pain associated with standing on a sloped surface. Gait Posture. 2013;37(3):313-8.

130. Khodaveisi H, Sadeghi H, Memar R, Anbarian M. Comparison of selected muscular activity of trunk and lower extremities in young women's walking on supinated, pronated and normal foot. Apunts Sports Med. 2016;51(189):13-9.

131. Menz HB, Dufour AB, Riskowski JL, Hillstrom HJ, Hannan MT. Foot posture, foot function and low back pain: the Framingham Foot Study. Rheumatology (Oxford). 2013;52(12):2275-82.

132. Powers CM, Chen PY, Reischl SF, Perry J. Comparison of foot pronation and lower extremity rotation in persons with and without patellofemoral pain. Foot Ankle Int. 2002;23(7):634-40.

133. Chuter V, Spink M, Searle A, Ho A. The effectiveness of shoe insoles for the prevention and treatment of low back pain: a systematic review and meta-analysis of randomised controlled trials. BMC Musculoskelet Disord. 2014;15:140.

134. Sadler S, Spink M, Lanting S, Chuter V. A randomised controlled trial investigating the effect of foot orthoses for the treatment of chronic nonspecific low back pain. Musculoskelet Care. 2023;21(3):856-64.

135. Bayıroğlu G, Pisirici P, Feyzioğlu Ö. The effect of different subtalar joint pronation amounts on postural stability, function and lower extremity alignment in healthy individuals. Foot. 2024;60:102123.

136. Dananberg HJ, Guiliano M. Chronic low-back pain and its response to custom-made foot orthoses. J Am Podiatr Med Assoc. 1999;89(3):109-17.

137. Nigg BM, Nigg S, Hoitz F, Subramanium A, Vienneau J, Wannop JW, et al. Highlighting the present state of biomechanics in shoe research (2000–2023). Footwear Sci. 2023;15(2):133-43.

138. Nakajima T, Sakamoto M, Tazoe T, Endoh T, Komiyama T. Location specificity of plantar cutaneous reflexes involving lower limb muscles in humans. Exp Brain Res. 2006;175(3):514-25.

139. Henry SM, Fung J, Horak FB. EMG responses to maintain stance during multidirectional surface translations. J Neurophysiol. 1998;80(4):1939-50.

140. Hoeffner R, Agergaard AS, Svensson RB, Cullum C, Mikkelsen RK, Konradsen L, et al. Tendon elongation and function after delayed or standard loading of surgically repaired Achilles tendon ruptures: a randomized controlled trial. Am J Sports Med. 2024;52(4):1022-31.

141. Agres AN, Arampatzis A, Gehlen T, Manegold S, Duda GN. Muscle fascicles exhibit limited passive elongation throughout the rehabilitation of Achilles tendon rupture after percutaneous repair. Front Physiol. 2020;11:746.

142. Frankewycz B, Penz A, Weber J, da Silva NP, Freimoser F, Bell R, et al. Achilles tendon elastic properties remain decreased in the long term after rupture. Knee Surg Sports Traumatol Arthrosc. 2018;26(7):2080-7.

143. Heikkinen J, Lantto I, Flinkkilä T, Ohtonen P, Niinimäki J, Siira P, et al. Soleus atrophy is common after the nonsurgical treatment of acute Achilles tendon ruptures: a randomized clinical trial comparing surgical and nonsurgical functional treatments. Am J Sports Med. 2017;45(6):1395-404.

144. Heikkinen J, Lantto I, Piilonen J, Flinkkilä T, Ohtonen P, Siira P, et al. Tendon length, calf muscle atrophy, and strength deficit after acute Achilles tendon rupture: long-term follow-up of patients in a previous study. J Bone Joint Surg Am. 2017;99(18):1509-15.

145. Winter DA. Biomechanics and motor control of human gait: normal, elderly and pathological. 2nd ed. Waterloo (ON): University of Waterloo Press; 1991.

146. Carmont MR, Gunnarsson B, Brorsson A, Nilsson-Helander K. Musculotendinous ruptures of the Achilles tendon have greater heel-rise height index compared with mid-substance ruptures with nonoperative management: a retrospective cohort study. J ISAKOS. 2024;9(2):148-52.

147. Srinivasan D, Mathiassen SE. Motor variability in occupational health and performance. Clin Biomech. 2012;27(10):979-93.

148. Davids K, Glazier P, Araújo D, Bartlett R. Movement systems as dynamical systems: the functional role of variability and its implications for sports medicine. Sports Med. 2003;33(4):245-60.

149. De Luca CJ. The use of surface electromyography in biomechanics. J Appl Biomech. 1997;13(2):135-63.

150. Dankaerts W, O'Sullivan PB, Burnett AF, Straker LM, Danneels LA. Reliability of EMG measurements for trunk muscles during maximal and submaximal voluntary isometric contractions in healthy controls and CLBP patients. J Electromyogr Kinesiol. 2004;14(3):333-42.

151. Mitchell MD, Yarossi MB, Pierce DN, Garbarini EL, Forrest GF. Reliability of surface EMG as an assessment tool for trunk activity and potential to determine neurorecovery in SCI. Spinal Cord. 2015;53(5):368-74.

152. Bogey R, Cerny K, Mohammed O. Repeatability of wire and surface electrodes in gait. Am J Phys Med Rehabil. 2003;82(5):338-44.

153. Camomilla V, Donati M, Stagni R, Cappozzo A. Noninvasive assessment of superficial soft tissue local displacements during movement: a feasibility study. J Biomech. 2009;42(7):931-7.

154. Zügner R, Tranberg R, Lisovskaja V, Shareghi B, Kärrholm J. Validation of gait analysis with dynamic radiostereometric analysis (RSA) in patients operated with total hip arthroplasty. J Orthop Res. 2017;35(7):1515-22.

155. Zügner R, Tranberg R, Lisovskaja V, Kärrholm J. Different reliability of instrumented gait analysis between patients with unilateral hip osteoarthritis, unilateral hip prosthesis and healthy controls. BMC Musculoskelet Disord. 2018;19(1):224.

156. Fändriks A, Zügner R, Shareghi B, Kärrholm J, Tranberg R. Skin and cluster markers underestimate knee flexion during controlled motions: evaluation of 12 patients with knee arthroplasty using radiostereometric analysis as reference. J Biomech. 2025;182:112591.

157. World Medical Association. Human experimentation: code of ethics of the World Medical Association. Declaration of Helsinki. Br Med J. 1964;2(5402):177.

158. ALLEA – All European Academies. The European Code of Conduct for Research Integrity [Internet]. Revised edition. Berlin: ALLEA; 2017. https://allea.org/code-of-conduct/

159. Larsson E, Brorsson A, Carling M, Johansson C, Carmont MR, Nilsson Helander K. Sex differences in patients' recovery following an acute Achilles tendon rupture: a large cohort study. BMC Musculoskelet Disord. 2022;23(1):913.

160. Larsson E, Nilsson N, Walstern J, Brorsson A, Nilsson Helander K. Females present larger deficit in heel-rise height at 3 months following an Achilles tendon rupture compared with males. Knee Surg Sports Traumatol Arthrosc. 2024;32(10):2581-88.

161. Alshabrami QM, Almansour BI, Al Shammari RA, Hameyd AY, Alabdelqader WF, Alenizy NF, et al. The use of physical therapy in managing sports injuries. J Int Crisis Risk Commun Res. 2024;7(S1-1):538.

162. Kerry R, Young KJ, Evans DW, Lee E, Georgopoulos V, Meakins A, et al. A modern way to teach and practice manual therapy. Chiropr Man Therap. 2024;32(1):17.

163. Tyni-Lenné R. To identify the physiotherapy paradigm: a challenge for the future. In: Andersson GBJ, Hobart DJ, editors. International perspectives in physical therapy. London: Taylor & Francis; 1989. p. 169-70.

164. Rowe R, Tichenor C, Bell S, Boissonnault W, King P, Kulig K. Orthopaedic manual physical therapy: description of advanced specialty practice. Tallahassee (FL): American Academy of Orthopaedic Manual Physical Therapists; 2008.

165. Lin I, Wiles L, Waller R, Goucke R, Nagree Y, Gibberd M, et al. What does best practice care for musculoskeletal pain look like? Eleven consistent recommendations from high-quality clinical practice guidelines: systematic review. Br J Sports Med. 2020;54(2):79-86.

166. Sharma P, Maffulli N. Biology of tendon injury: healing, modeling and remodeling. J Musculoskelet Neuronal Interact. 2006;6(2):181-90.

15. Appendix

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> **My coach said I run like a girl, and I said if he ran a little faster, he could too.**
>
> MIA HAMM

Sahlgrenska University Hospital – Occupational Therapy and Physiotherapy

Implementation / Physiotherapeutic Treatment Plan Following Acute Achilles Tendon Rupture
Applicable to both surgical and non-surgical management

Emergency Department (Initial Management):

The affected foot is immobilized in a cast in plantarflexion.

Provide the patient with the brochure “Information for Patients with Achilles Tendon Injury” and review the prescribed exercises.

Advise the patient to avoid placing the foot on the floor whenever possible; the foot should be kept elevated in a horizontal position during rest to minimize swelling.

When standing (e.g., for hygiene, dressing, or food preparation), the foot may rest lightly on the floor for balance.

The patient should off-load the injured leg using two crutches.

Gait should involve active hip and knee movement on the affected side without weight-bearing on the foot.

Patients with impaired balance may require a walking aid (e.g., Beta crutch), and in some cases, a wheelchair. Toe-touching the ground with the injured foot may aid in balance.

2 Weeks Post-Injury:

At the orthopedic outpatient follow-up, the cast is replaced with an orthosis, which must be worn at all times, including during showering.

Instruct in the home exercise program “Achilles Tendon Program I.”

Continue to encourage toe mobility.

Educate the patient to routinely palpate the tendon to monitor for signs of overload (increased warmth, redness, swelling).

Teach how to remove and reapply the orthosis for hygiene and exercise purposes.

Encourage regular ventilation of the lower leg and provide instructions on cleaning the foot and leg with a washcloth.

Fit the orthosis with three wedges as appropriate (typically smaller wedges). Patients may add a personal insole or silicone heel cup for comfort.

If tension is felt in the tendon during gait with three wedges, a fourth wedge may be added temporarily and removed by the patient after several days based on comfort.

Begin full weight-bearing gait training with two crutches.

Practice gait on flat surfaces and stairs if applicable, ensuring no uncomfortable stretching of the tendon.

Recommend a cork wedge in the contralateral shoe to equalize leg length. For patients with special needs, refer to an orthopedic technician for shoe modification.

Begin planning for appropriate footwear for use post-orthosis. Indoor and outdoor shoes should have a heel height differential of at least 1.5 cm. Flat shoes (e.g., sneakers) are not appropriate; both pairs should have a heel counter. Shoes can be assessed at the next visit.

Plan for continued physiotherapy. If referred to primary care, ensure the receiving provider is aware of their role in early rehabilitation and responsible for progressive wedge and orthosis weaning, starting at 8 weeks post-injury.

4 Weeks Post-Injury:

Follow-up with treating physiotherapist.

Review “Achilles Tendon Program I.”

Assess the tendon: continuity, width, consistency.

Perform the Thompson test.

Evaluate current tendon loading; if needed, reduce walking volume.

Remove the lowest wedge.

Retrain gait on flat and stair surfaces.

Wean off crutches indoors if gait is well-balanced and the tendon is not under excessive strain.

Assess footwear. Heel elevation should remain ≥1.5 cm. Shoes must have a heel counter.

Determine frequency and location of continued rehab.

Surgical patients: orthosis can be removed at night and for seated showering (non-weight-bearing).

Non-surgical patients: orthosis must remain on 24/7.

For surgical cases: monitor wound healing and begin scar mobilization once healed.

All patients may initiate stationary cycling with the orthosis on. Provide training in safe mounting/dismounting.

Surgical patients should avoid excessive sweating in the orthosis until the wound is fully healed.

6 Weeks Post-Injury:

Follow-up with physiotherapist.

Repeat assessments: tendon palpation, Thompson test, and loading evaluation.

Remove the next wedge (done by the patient or physiotherapist).

Continue gait training. If discomfort arises, resume use of two crutches.

The final wedge remains until orthosis discontinuation at 8 weeks post-injury.

8 Weeks Post-Injury:

Follow-up and orthosis weaning.

Repeat tendon assessment, Thompson's test, and evaluate loading.

Measure Achilles Tendon Resting Angle (ATRA).

For surgical patients: continue scar evaluation and mobilization. Provide tape application guidance for scar maturation.

Begin "Achilles Tendon Program II." Seated exercises are performed barefoot.

During standing/walking, indoors and outdoors, supportive shoes must be worn for four weeks post-orthosis.

Distribute cork wedges for use in both shoes.

Emphasize gait training with appropriate foot roll-off.

Restoration of normal gastrocnemius activation is key to reducing pain, swelling, and optimizing function.

Each step should promote natural tendon loading and circulation.

Aim for symmetry in step length, stance phase, and muscle activation across joints.

Continue using one or two crutches until gait is normalized.

Identify and discuss risk factors (e.g., stairs, curbs, inclines, uneven terrain).

Advise against walking as exercise; cycling is preferred to stimulate circulation. Compression stockings may also be beneficial.

Driving should be avoided until coordination and strength are sufficient.

8–12 Weeks Post-Injury:

Ongoing rehabilitation with gradual progression of load and speed, based on individual assessment.

Focus on restoring automatic, normalized gait pattern.

Crutches should be used until gait mechanics are optimal.

Balance protection of healing structures with appropriately timed loading.

Many patients benefit from more frequent physiotherapy visits following orthosis discontinuation.

At 12-week orthopedic follow-up, include completed ATRS (Achilles Tendon Total Rupture Score) and activity level (PAS, Grimby scale).

12 Weeks and Beyond:

Continue individualized rehabilitation. Progressively increase load with the goal of returning to pre-injury function.

Stretching of the gastrocnemius-soleus complex should be avoided before 16 weeks post-injury, as elongation is a major risk for compromised recovery.

For patients with ankle stiffness, introduce range-of-motion exercises in positions that minimize Achilles tendon stretch (e.g., knee flexed, foot supported).

Outdoor cycling can typically begin after 12 weeks, starting cautiously.

Total rehabilitation time ranges from 6–12 months.

Sport-specific training may begin at 4–6 months if progress is satisfactory.

Return to high-impact sport typically requires 9–12 months.

Full tendon tensile strength is generally not regained until 12 months post-injury.

Foot – Achilles Tendon Program I
Home Exercise Program
Sahlgrenska University Hospital – Occupational Therapy and Physiotherapy

Instructions:
During the first 6 weeks while wearing the orthosis, remove it 3–5 times daily to perform these exercises.

Important Considerations During Rehabilitation Following Achilles Tendon Rupture:

There are two key risks that must be avoided:

Re-rupture of the tendon

Tendon healing with excessive length, which leads to decreased strength

To minimize these risks:

Avoid excessive stretching of the tendon in the early phase.

Stretching occurs when the foot is dorsiflexed (bends upward).

For the tendon to heal optimally and regain elasticity, gentle and appropriate activation is essential.

Example set-up: Orthosis on the left foot, with 3 wedges placed under the grey insole.

Exercise 1

Ankle plantarflexion in prone or sitting

Position:

Lying face down with your feet hanging off the edge of a bed, or

Sitting with the lower leg supported and knee straight

Action:

Point the foot down (plantarflex), then slowly return to a relaxed position.

Dosage:
3 sets of 20 repetitions

Exercise 2

Side-to-side ankle movement

Position:

Seated or lying face down as in Exercise 1

Action:

Gently move the foot from side to side in a small arc.

Keep the knees still.

Dosage:
3 sets of 20 repetitions

Exercise 3

Supported seated heel raise in orthosis

Position:

Seated, with orthosis on

Deflate the orthosis air cushions and open the front panel

Hold the top of the orthosis with both hands for support

Action:

Perform a small heel lift, keeping the forefoot in contact with the sole.

Do not grip with the toes.

This safely activates the calf muscle while the wedges protect the tendon from stretching.

Dosage:
2 sets of 10 repetitions

General Guidelines:

All exercises should be comfortable and pain-free - no stretching, tingling, or sharp pain.

Always wear the orthosis when standing or walking.

It’s normal for the tendon and ankle area to be slightly warmer and more swollen than the uninjured side during healing.

However, progressive swelling, redness, or increasing warmth may indicate overuse of the tendon and should be addressed.

Foot – Achilles Tendon Program II
Home Exercise Program
Sahlgrenska University Hospital – Occupational Therapy and Physiotherapy

Instructions:

Perform the exercises **3–5 times per day**.

Circulation

1) Gentle ankle pumps

Pump the foot up and down softly.

You may place a **roll under the knees** for support.

The movement should be **pain-free** and **comfortable** for the tendon.
20 repetitions

Activation

2) Foot cupping with relaxed toes

Gently form a dome shape with the foot while keeping the toes relaxed.
10 repetitions

Mobility

3) Forward-backward weight shift

Keep the entire foot sole in contact with the floor.

Move the knee and body weight forward and backward over the foot.

On the final repetition, hold the end position for a few seconds.
10 repetitions

Strength

4) Seated heel raise

Sit with the **entire forefoot on the floor**.

You may rest your hands on your knees.

Lift the heels to perform a **heel raise**.
20 repetitions

Seated foot jogging

Perform small alternating bouncing movements with the feet while seated.
20 repetitions

Heel raise

Repeat the seated heel raise again.
20 repetitions

Standing Exercises (with shoes)

Instructions: Perform standing exercises with **shoes on**.

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5) Lateral weight shifting (side-to-side)

Transfer weight from one leg to the other.

Load the entire foot on the stance leg.

Feel muscle activation, especially in the **gluteal region**.
10 repetitions

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6) Gait preparation exercise

Stand with a **book or rolled towel under both heels** and lean against a wall.

Alternate lifting and lowering each heel gently.

Focus on **stable hips**, **relaxed toes**, and **fluid knees**.
20 repetitions

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7) Gait training

Walk in a straight line, placing one foot directly in front of the other.

Imagine walking on a **tightrope**.

Walk **10 steps forward**, then **10 steps backward**.

Use **crutches until you feel stable without them**.

Walking Guidelines:

Aim to walk with a normal gait pattern: heel strike, roll through, and even step length.

Short steps are often better in the early phase.

Use crutches as long as needed.

Walk as much as necessary in daily life, but **do not walk for exercise** at this stage.

Many post-injury issues are caused by **over-walking** during early recovery.

To Reduce or Prevent Swelling:

Elevate the foot while sitting.

Perform **seated ankle pumps** and **toe spreading and gripping**.

If needed, consider using a **compression sock**.