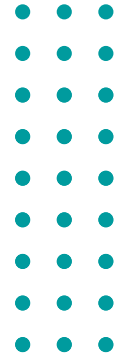


Biomechanical evaluation of a hip brace for the treatment of mild-to-moderate hip osteoarthritis



Hannah Sophie Steingrebe

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of mild-to-moderate hip osteoarthritis

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Summary

Hip osteoarthritis (HOA) is a highly prevalent musculoskeletal disorder characterised by progressive cartilage degeneration, subchondral bone remodelling, and synovial inflammation, causing joint pain and stiffness and leading to reduced mobility and quality of life in affected individuals.

While the prevention of initial HOA development would be the ultimate treatment goal to overcome the individual and socioeconomic burden of this disease, current therapy approaches focus on disease deceleration and treatment of symptoms. Following the progressive nature of HOA, therapy usually commences with conservative treatments, such as exercise or weight reduction, for individuals with mild symptoms, continues with pharmacological pain management and ultimately requires the partial or total surgical replacement of the hip joint. However, well-investigated and effective conservative treatment options for HOA are sparse. This lack of research also pertains to the use of hip braces for the treatment of HOA. Some studies have demonstrated beneficial effects of bracing on hip pain and joint function for individuals with HOA, but a comprehensive understanding of the mechanisms underlying these bracing effects is missing.

The hip joint, as the connection between the upper and lower body, serves two essential functions during human bipedal locomotion: it has to carry and stabilise the torso while simultaneously allowing sufficient joint mobility.

Numerous studies have demonstrated impaired locomotion in individuals with HOA, characterised by lower movement velocity, reduced joint mobility, and lower joint moments and contact forces. A great portion of research has focused on individuals with end-stage HOA, while locomotion of individuals with mild or moderate symptoms has been studied less frequently. Moreover, analyses of locomotion are mostly limited to level walking.

This thesis extends the knowledge on the impact of mild-to-moderate HOA on locomotion by comprehensively evaluating its effect on joint biomechanics during various locomotion tasks, postural control, joint proprioception and overall physical function.

Based on these insights, the effectiveness of an elastic hip brace for treating mild-to-moderate HOA is explored by studying the effects on the aforementioned locomotion characteristics and pain.

To lay the foundation for evaluation of the hip brace, firstly, a systematic literature review and meta-analysis were conducted, secondly, a method for measuring hip proprioception was developed, and thirdly, the test-retest reliability of this method was assessed.

The effects of mild-to-moderate HOA and the effectiveness of hip bracing were then assessed in a combined case-control and intervention study, including 21 participants with unilateral mild-to-moderate symptomatic HOA and 21 matched healthy control participants. Biomechanical motion analysis was performed during level walking, steady-state stair walking and turning. Hip proprioception, functional capacity and hip pain were assessed using an joint position reproduction test, the 6-Minute-Walking-Test, and visual analogue scales, respectively. In doing so, not only immediate bracing effects but also effects after one week of bracing were evaluated.

The study confirmed previously described effects of HOA on locomotion biomechanics in individuals with moderate-to-severe HOA, including reduced locomotion speed, lower hip extension, adduction, and internal rotation angles, as well as lower hip joint moments. However, effects are less pronounced in individuals with mild-to-moderate HOA than in those with end-stage HOA. By analysing more demanding locomotion tasks, e.g. stair ascent and turning, HOA-induced impairments of locomotion can be identified at an earlier disease stage, allowing for the timely initiation of therapy.

Wearing the elastic hip brace resulted in a reduction in hip pain and an increase in functional capacity and locomotion velocity. Moreover, bracing induced changes in sagittal hip and pelvis motion, indicating improved pelvis mobility, which is likely responsible for the observed increase in locomotion velocity. Furthermore, brace-induced restriction of peak hip adduction and internal rotation may reduce hip pain and joint loading, but could also lead to decreased muscle activity. During some locomotion tasks, bracing caused higher joint moments at the hip joint and, to a lesser extent, at the knee joint. While this counteracts the lower joint moments observed in individuals with HOA, higher joint moments likely require higher muscle forces, and the consequences for joint loading remain unclear.

Future studies need to investigate the effects of long-term brace use on muscle force and joint loading. The observed bracing effects depended on the analysed locomotion task, and results suggested different modes of action for the different movement planes of the hip joint. By including a retention analysis in future studies, the understanding of the underlying causes of the observed brace-induced changes in locomotion biomechanics could be further improved.

Zusammenfassung

Hüftarthrose (HOA) ist eine weitverbreitete Muskel-Skelett-Erkrankung, die durch eine progressive Zerstörung des Gelenkknorpels, pathologische Veränderungen im subchondralen Knochengewebe und entzündliche Prozesse in der Gelenkkapsel charakterisiert ist. Dabei führt HOA zu Schmerzen und Steifheit des Gelenks und schränkt so die Mobilität der betroffenen Personen ein und vermindert ihre Lebensqualität.

Um die schweren individuellen und sozioökonomischen Folgen von HOA zu bekämpfen, müsste im Idealfall der initialen Entstehung der HOA vorgebeugt werden. Aktuell zielen die meisten Therapieansätze jedoch auf eine Verlangsamung des Fortschreitens und eine Behandlung der Symptome ab. Kongruent zum progressiven Fortschreiten der Erkrankung ist auch die Therapie für HOA stufenweise aufgebaut. Während im Anfangsstadium zunächst konservative Behandlungsmethoden wie z. B. Krafttraining oder Gewichtsreduktion zum Einsatz kommen, wird mit zunehmendem Fortschreiten der Einsatz pharmakologischer Therapien notwendig, um die Symptome wie Schmerz zu behandeln. Im schwergradigen Krankheitsstadium wird schließlich ein partieller oder totaler Gelenkersatz nötig. In der Therapie von HOA mangelt es dabei vor allem an gut untersuchten und effektiven konservativen Behandlungsmethoden. Dieser Mangel an Forschung betrifft auch die Nutzung von Hüftorthesen in der Behandlung von HOA. Bisher konnte in den wenigen verfügbaren Studien zum Nutzen von Hüftorthesen ein positiver Effekt auf das Schmerzempfinden und die allgemeine körperliche Funktion gezeigt werden. Jedoch fehlt ein umfassendes Verständnis der durch Orthesen hervorgerufenen Effekte und der zugrunde liegenden Wirkmechanismen.

Das Hüftgelenk spielt als Verbindung zwischen dem Oberkörper und den Beinen eine wichtige Rolle bei der menschlichen Fortbewegung. Die Anatomie des Hüftgelenks ist daher durch seine zwei Hauptfunktionen gekennzeichnet: hohe Stabilität zum Tragen und Stabilisieren des Oberkörpers bei gleichzeitig größtmöglicher Mobilität für die Fortbewegung.

Eine Vielzahl an Studien hat gezeigt, dass die Fortbewegung durch das Auftreten von HOA beeinträchtigt ist. So wurden z. B. eine niedrigere Bewegungsgeschwindigkeit, Gelenkbeweglichkeit sowie verminderte Gelenkmomente und Gelenkkontraktkräfte festgestellt.

Viele dieser Befunde wurden jedoch in Kohorten von Personen mit schwergradiger HOA gefunden und Ergebnisse zu Personen mit milder oder moderater HOA sind rar. Weiterhin

wurde zumeist die Beeinträchtigung von ebenem Gehen durch HOA untersucht, während Ergebnisse zu anderen Lokomotionsbewegungen wie z. B. dem Treppensteigen fehlen.

Die vorliegende Dissertation erweitert die Erkenntnisse des Einflusses von milder bis moderater HOA auf die menschliche Lokomotion durch eine umfassende Analyse der Effekte auf die Gelenkbiomechanik während verschiedener Lokomotionsbewegungen, die posturale Kontrolle, die Hüftpropriozeption sowie die allgemeine Gelenkfunktion.

Basierend auf den gewonnenen Erkenntnissen wurde die Effektivität einer elastischen Hüftorthese für die Behandlung von milder bis moderater Hüftarthrose, auf Basis der zuvor genannten Parameter sowie des Schmerzempfindens, evaluiert. Dabei wurden nicht nur die Effekte, welche unmittelbar nach Anpassung der Orthese auftreten untersucht, sondern auch die Effekte nach einer einwöchigen Tragephase analysiert.

Als Grundlage für die wissenschaftliche Evaluation der Hüftorthese wurde zunächst eine systematische Literaturanalyse und Meta-Analyse zum Einfluss von HOA auf die Kinematik von Lokomotionsbewegungen durchgeführt. Weiterhin wurde ein Messverfahren zur Erfassung der Hüftpropriozeption entwickelt und die Test-Retest-Reliabilität des Verfahrens evaluiert. Der Einfluss von milder bis moderater Hüftarthrose sowie die Effektivität der Behandlung mit einer Hüftorthese wurden anschließend in einer kombinierten Fall-Kontroll- und Interventions-Studie in einer Kohorte von 21 Personen mit milder bis moderater HOA sowie 21 gesunden Kontrollpersonen untersucht.

In der Studie wurde eine biomechanische Bewegungsanalyse des Gehens, Treppensteigens sowie Kurvengehens durchgeführt. Weiterhin wurde die Hüftpropriozeption mittels eines Winkel-Reproduktions-Tests, die allgemeine Gelenkfunktion mittels des 6-Minuten-Gehtests sowie das Schmerzempfinden mittels visueller Analogskalen erfasst.

Die Ergebnisse der Studie bestätigen frühere Studienergebnisse zur Beeinflussung der Lokomotion durch moderate oder schwergradige HOA wie z. B. eine niedrigere Bewegungsgeschwindigkeit, reduzierte Gelenkwinkel in der Hüftextension, -adduktion und -innenrotation sowie verminderte Gelenkmomente am Hüftgelenk. Jedoch sind die Effekte bei Personen mit milder bis moderater HOA weniger stark ausgeprägt. Durch die Analyse von anspruchsvollen Bewegungsaufgaben wie z. B. dem Treppensteigen oder Kurvengehen können HOA-bedingte Einschränkungen der Lokomotion frühzeitig identifiziert werden, um so einen frühzeitigen Behandlungsbeginn zu ermöglichen.

Das einwöchige Tragen der Orthese führte zu einer Verminderung des Hüftschmerzes sowie einer Erhöhung der funktionellen Kapazität und Bewegungsgeschwindigkeit. Des Weiteren kam es durch das Tragen der Orthese zu Veränderungen in der sagittalen Becken- und Hüft-

kinematik, was auf eine verbesserte Mobilität des Beckens hindeutet und die Erhöhung der Bewegungsgeschwindigkeit ermöglicht. Weiterhin führte das Tragen der Orthese zu einer Verminderung der maximalen Adduktions- und Innenrotationswinkel im Hüftgelenk, was ggf. zur beobachteten Schmerzreduktion geführt haben könnte, jedoch auch die Muskelaktivität und folglich die Gelenkbelastung am Hüftgelenk reduzieren kann. Mit der Orthese wurden höhere Gelenkmomente am Hüftgelenk und teilweise auch am Kniegelenk beobachtet und das Tragen der Orthese wirkt damit der beobachteten Reduzierung der Gelenkmomente bei Personen mit HOA entgegen. Den höheren externen Gelenkmomenten müssen folglich auch höhere durch Muskeln generierte interne Momente entgegenwirken, was wiederum Einfluss auf die Gelenkbelastung haben könnte.

Aus diesem Grund bedarf es weiterführender Studien, welche die ortheseninduzierten Effekte auf die Muskelaktivität, Muskelkraft und Gelenkbelastung mit besonderem Fokus auf die langfristige Nutzung von Hüftorthesen analysieren. Während die beobachteten Effekte zum einen abhängig von der Lokomotionsbewegung sind, zeigte die vorliegende Arbeit andererseits Hinweise auf unterschiedliche Wirkmechanismen in den verschiedenen Bewegungsebenen des Hüftgelenks. Die Inklusion einer Retentionsbedingung in zukünftigen Studien könnte hier das Verständnis der den beobachteten biomechanischen Effekten zugrunde liegenden Wirkmechanismen der Hüftorthese weiter vertiefen.

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List of Abbreviations

ANCOVA	Analysis of covariance
ASIS	Anterior superior iliac spine
BMI	Body Mass Index
BoS	Base of support
CoM	Centre of mass
CON	Control
FAI	Femoroacetabular impingement
GC	Gait cycle
GMed	Gluteus medius
GRF	Ground reaction force
HA	High angle
HHS	Harris Hip Score
HJC	Hip joint centre
HOA	Hip osteoarthritis
HOOS	Hip Osteoarthritis Outcome Score
IC	Initial contact
IfSS	Institute of Sports and Sports Science
IMU	Inertial measurement unit
JPR	Joint position reproduction
JPS	Joint position sense
K-L	Kellgren-Lawrence
KIT	Karlsruhe Institute of Technology
KOA	Knee osteoarthritis
LA	Low angle
MD	Mean difference
MoS	Margin of stability
MWU	Mann-Whitney-U test
OA	Osteoarthritis
OHS	Oxford Hip Score
PRISMA	Preferred Reporting Items for Systematic Reviews and Meta-Analyses
PROSPERO	International Prospective Register of Systematic Reviews
PSIS	Posterior superior iliac spine
RoM	Range of motion
SD	Standard deviation
SIJ	Sacroiliac joint
SMD	Standardised mean difference
SnPM	Statistical nonparametric mapping
SPM	Statistical parametric mapping

List of Abbreviations

ST	Stance phase
THR	Total hip replacement
TO	Toe-off
TUG	Timed Up and Go
WCX	Wilcoxon test
WOMAC	Western Ontario and McMaster Universities Osteoarthritis Index
6MWT	6-Minute-Walking-Test
95%CI	95 % confidence interval

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1 Introduction

1.1 Motivation

Increasing prosperity, progress in medicine and hygiene, better education, as well as improved work and life conditions, have led to a dramatic increase in life expectancy (Christensen & Vaupel, 1996). The downside of this development is a drastic increase in age-related musculoskeletal disorders such as hip osteoarthritis (HOA) (Scheuing et al., 2023). Approximately one-quarter of individuals reaching the age of 85 experience symptomatic HOA (L. Murphy et al., 2010), meaning that they suffer from pain and loss of function (Cibulka & Threlkeld, 2004). Especially the impairment of gait ability, as one of the most fundamental human movements crucial for participation in social and professional life, causes a substantial loss of quality of life (Vogt, 2005). What options do we have to improve pain, function, and quality of life for individuals affected by this condition? Hip osteoarthritis develops progressively, and so do the therapy options. Starting with conservative, non-pharmacological treatments for mild-to-moderate HOA, disease progression at some point requires the intake of pharmaceuticals, and finally, the application of surgical procedures, such as total hip replacement (THR) (Dieppe & Lohmander, 2005). However, while the use of pharmaceuticals and surgical treatments has been extensively studied in the past decades (Ferguson et al., 2018; Richard et al., 2023), knowledge of conservative non-pharmacological treatments remains sparse and is often extrapolated from research on knee osteoarthritis (KOA) (Bannuru et al., 2019). Currently, recommendations for conservative treatment are limited to (mind-body) exercise, weight loss and cane use (Kolasinski et al., 2020). Another conservative treatment method is the application of braces. While the American College of Rheumatology recommends bracing for persons with KOA (Kolasinski et al., 2020), no recommendations were made for bracing in HOA due to a lack of research. Expanding the range of non-pharmacological conservative treatments could help reduce both the individual and socioeconomic burden of this highly prevalent disease. Therefore, this thesis aims to comprehensively assess the effects of hip bracing for the treatment of mild-to-moderate HOA.

1.2 Outline of the thesis

The present thesis consists of twelve chapters. Following this introduction, background information is given, and the current state of research is reviewed (chapter 2), leading to the aim and scope of this thesis (chapter 3). The seven subsequent chapters represent independent manuscripts which are based on one systematic literature review and meta-analysis, as well as two individual studies (Table 1.1). Thereby, manuscripts focus on (a) preliminary research questions such as the reliability of a testing procedure for the assessment of hip proprioception (chapter 5) or the evaluation of hip mobility required for different locomotion tasks (chapter 6), (b) the effects of mild-to-moderate HOA on functional capacity, hip proprioception, postural control and locomotion biomechanics (chapters 4, 7, 8, 9) and (c) the effect of hip bracing on pain, functional capacity, hip proprioception, and locomotion biomechanics (chapters 8, 9, 10). Some of the manuscripts cover two of the categories mentioned above (Table 1.1). Chapter 11 contains a comprehensive discussion of the results with regard to existing literature and future research directions. Lastly, an overall conclusion is drawn in chapter 12.

Table 1.1: Scheme of this dissertation project with the corresponding published manuscripts.

Preliminary work	Effects of hip osteoarthritis on ...	Effects of hip bracing on ...
Reliability of hip joint position sense measurement (Steingrebe et al., 2019 – ECSS 2018 Book of Abstracts)	... locomotion kinematics - systematic review and meta-analysis (Steingrebe et al., 2023 - <i>Frontiers in Sports and Active Living</i>)	
Hip mobility requirements and joint loading during various locomotion movements (Steingrebe et al., 2024 – <i>Journal of Clinical Medicine</i>)	... level walking biomechanics, functional capacity, proprioception and pain (Steingrebe et al., 2022 - <i>Frontiers in Bioengineering and Technology</i>)	
	... turning biomechanics (Steingrebe et al., 2025 - <i>Clinical Biomechanics</i>)	
	... stair walking biomechanics and postural control (Steingrebe et al., 2025 - <i>Journal of Biomechanics</i>)	... stair walking biomechanics and pain (currently under review - <i>BMC Musculoskeletal Disorders</i>)

Note. ECSS = European College of Sport Science

All but two manuscripts have been published in international peer-reviewed journals. The manuscript in chapter 5 is an unpublished extended version of an abstract from the 2019 European College of Sport Science Congress, published in the book of abstracts (Steingrebe et al., 2019), while the manuscript in chapter 10 is currently under review for publication in BMC Musculoskeletal Disorders.

2 Theoretical background

2.1 Definition and classification of osteoarthritis

Historically, osteoarthritis (OA) has been perceived as a “wear-and-tear” condition that develops with age and progressively damages the joint cartilage, consequently limiting the margin for therapeutic interventions (Ehrlich, 2003). However, with increasing research, it has become evident that OA does not solely involve progressive cartilage deterioration but rather affects all structures of a joint (Block & Shakoor, 2009). Therefore, more comprehensive definitions of OA state that it is “[...] a painful degenerative process involving all structures of involved joints, that is not primarily inflammatory, and which involves progressive deterioration of articular cartilage and reactive alterations of subchondral bone and surrounding joint structures” (Block & Shakoor, 2009). Joint cartilage degeneration and remodelling of subchondral bone can be visualised with radiography, e.g. by means of the joint space width or the presence of osteophytes. Thereby, the most common classification for radiographic OA is the Kellgren-Lawrence (K-L) grade (Fan et al., 2023), which includes scores from zero to four interpreted as 0 = no presence of OA; 1 = doubtful presence of OA; 2 = undoubted presence of OA with minimal severity; 3 = OA with moderate severity and 4 = OA with severe severity (Kellgren & Lawrence, 1957).

Yet, further research has shown that radiographic disease symptoms, e.g. joint space narrowing, are not highly correlated with illness symptoms like joint pain, and often radiographic signs of OA are found in joints of persons without symptoms (Buckwalter et al., 2001; Kraus et al., 2015). Based on this observation, Kraus et al. (2015) consider radiographic disease features as risk factors for the development of illness and propose a novel OA definition that stresses the distinction between OA as a disease and OA as an illness:

Osteoarthritis is a disorder involving movable joints characterized by cell stress and extracellular matrix degradation initiated by micro- and macro-injury that activates maladaptive repair responses including pro-inflammatory pathways of innate immunity. The disease manifests first as a molecular derangement (abnormal joint tissue metabolism) followed by anatomic, and/or physiologic derangements (characterized by cartilage degradation, bone remodeling, osteophyte formation, joint

inflammation and loss of normal joint function), that can culminate in illness. (Kraus et al., 2015)

Consequently, classification of OA should be based on clinical and radiographic criteria, as this provides higher sensitivity and specificity (Altman et al., 1991). The most widely used clinical criteria for OA are those of the American College of Rheumatology, which include, besides other criteria, overall hip pain combined with pain on internal rotation of the hip, limited hip internal rotation range of motion (RoM) ($< 15^\circ$) and morning stiffness of 60 minutes or less (Altman et al., 1991).

2.2 Prevalence and burden of hip osteoarthritis

Besides the knee and hand joints, the hip joint is most commonly affected by OA (Oliveria et al., 1995). However, based on the various definitions used for HOA assessment (radiographic, symptomatic, or self-reported) (Pereira et al., 2011) and the characteristics of the studied sample (Bennell, 2013), the prevalence of HOA varies strongly. While the overall prevalence of radiographic HOA is 8.6 %, regional differences are large, with the lowest prevalence of 1.2 % found in Africa and the highest prevalence of 12.6 % found in Europe (Fan et al., 2023). Moreover, the prevalence of HOA increases with age (Bijlsma & Knahr, 2007; Fan et al., 2023). In African-American or Caucasian individuals aged 45 or above, prevalence rates of 27.6 % for radiographic HOA, 9.7 % for symptomatic HOA and 2.5 % for severe radiographic HOA have been reported (Jordan et al., 2009). Furthermore, the lifetime risk of developing HOA is approximately 25 % at the age of 85 (L. Murphy et al., 2010).

For the affected persons, HOA has serious consequences on the quality of life due to pain and loss of function (Dekker et al., 2009; Salaffi et al., 2005). Pain is mainly perceived at the groin and buttocks, but also below the ipsilateral knee joint (Khan et al., 2004). Initially, perceived pain is mostly triggered by an unfamiliar or dynamic activity (e.g. sport), and eventually, affected individuals refrain from high-impact activities (e.g. skiing). In advanced stages, the pain becomes constant and impairs activities of daily living (e.g. stair walking). In severe HOA, constant pain and short episodes of unpredictable, intense pain are experienced, leading to avoidance of activities, including social and recreational activities. Thereby, the increase from mild pain to severe pain is more rapid in individuals with HOA than in those with KOA (Hawker et al., 2008), and 50 % of individuals with HOA undergo hip replacement surgery within 2.5 years after initial consultation (Dabare et al., 2017).

Total hip replacement surgeries can provide pain relief and restore functional mobility (Neuprez et al., 2020), and therefore have gained increasing popularity over the last decades. The number of THR surgeries increased by 22 % between 2009 and 2019 (OECD, 2021), and a lifetime risk of about 10 % exists to undergo THR for the treatment of HOA (Culliford et al., 2012). As the expected human lifespan increases, the number of individuals affected by HOA and consequently the amount of THR surgeries is projected to increase (Fuchs et al., 2017; Kurtz et al., 2007). If this trend continues, the socioeconomic burden of HOA might become unsustainable (Ackerman et al., 2019).

This stresses the need for effective conservative treatment methods that are able to reduce pain and improve function and quality of life in order to reduce the number of primary THR or revision surgeries.

2.3 Anatomy of the hip joint and hip osteoarthritis

The hip joint plays a crucial role in locomotion as the articulation between the lower and upper body (Sangeux, 2019). Consequently, the anatomy of the hip joint is characterised by two main goals: maintaining stability to carry the torso and simultaneously allowing the greatest possible mobility to facilitate locomotion (Zimmer & Appell, 2021, p. 75). The hip is a ball-and-socket joint with the concave-shaped acetabulum, formed by the os ischium, os ilium and os pubis, and the spherical femoral head as joint partners. High congruency of the joint partners leads to a large contact area (Hamill et al., 2015, p. 178). The joint surface is additionally enlarged by the acetabular labrum, which is attached to the rim of the acetabulum (Byrne et al., 2010). As the acetabulum and the labrum cover the femoral head beyond its equator, the joint is defined as an enarthrosis spherioidea (Wurzing, 2007, p. 328). Consequently, in contrast to e.g. the knee joint, a high degree of bony stability is obtained, which is complemented by a strong and dense joint capsule further reinforced by three strong extra-capsular ligaments: the iliofemoral ligament, the ischiofemoral ligament and the pubofemoral ligament. All three ligaments, to some extent, hinder excessive hip extension (Wurzing, 2007, p. 333), as in hip extension combined with slight hip abduction and internal rotation, all parts of the joint capsule are tight (Lloyd-Roberts, 1953). Therefore, shortening of any aspect of the capsule hinders extension of the hip joint (Lloyd-Roberts, 1953), and reduced elasticity of the joint capsule is observed in hips with severe HOA (Tara-sevicius et al., 2007).

The femoral head is attached to the more laterally positioned femur shaft by the femoral neck (Figure 2.1). This structure facilitates free motion of the femoral head within the acetabulum and increases the lever arm of the hip abductor muscles (Byrne et al., 2010). The angle between the femur shaft and neck (neck-shaft angle) decreases from about 150° at birth to $125 \pm 5^\circ$ in adults. Large neck-shaft angles of more than 130° (termed coxa valga) or small neck-shaft angles below 120° (termed coxa vara) impact the lever arm of the abductor muscles. The femoral anteversion, which describes the anterior rotation of the femoral neck, is about 15 to 20° (Byrne et al., 2010). This anteversion enhances the contact between the femoral head and the anterior superior portion of the acetabulum and increases the lever arms of the Mm. gluteus maximus and piriformis (Anderson & Blake, 1994). Increased neck-shaft and anteversion angles have been shown to cause increased contact forces in the hip joint during walking (Kainz et al., 2023), and increased femoral anteversion has been linked to increased HOA risk (Parker et al., 2021; Tiburzi et al., 2024).

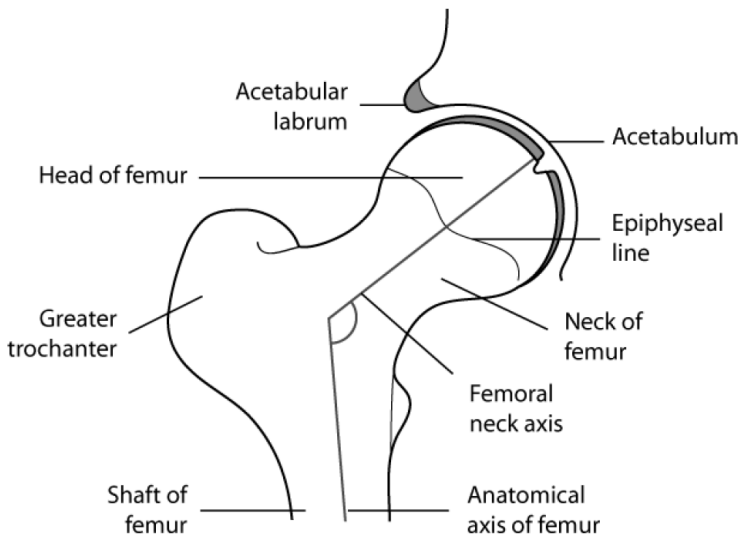


Figure 2.1: Cross-sectional view of the normal hip joint (Byrne et al., 2010).

The hip joint possesses three degrees of freedom (Hamill et al., 2015, p. 178), allowing movement in three planes of motion: flexion-extension (sagittal plane), abduction-adduction (frontal plane), and internal-external rotation (transverse plane). Passive joint mobility is largest in the sagittal plane with up to 140° of flexion and 15° of extension (with concurrent extension of the knee joint). In 90° of hip flexion, internal rotation angles of 35° and external rotation angles of 45° are possible. In the frontal plane, mobility is larger in abduction, with

approximately 40°, and only 25° of adduction (Wurzing, 2007, p. 334). However, lower passive hip RoM in all movement directions has been described in individuals with HOA (M. Arokoski et al., 2004; Baker et al., 2016; Klässbo et al., 2003).

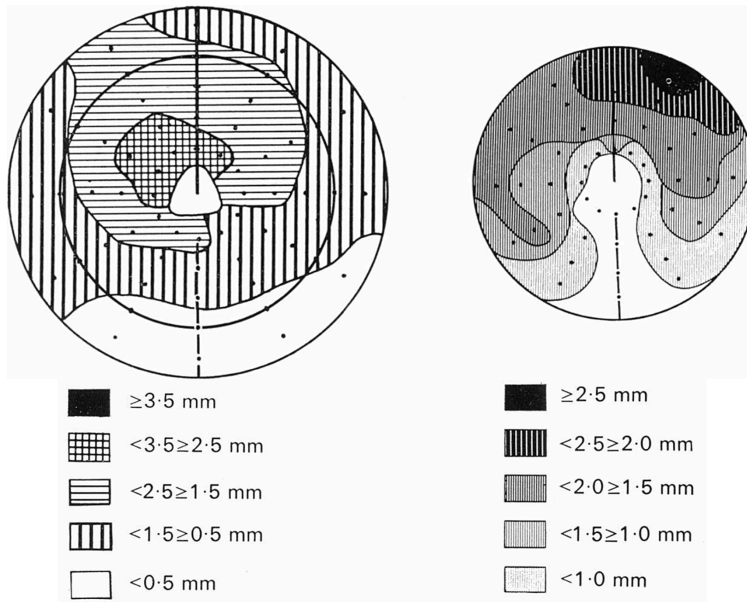


Figure 2.2: Average cartilage thickness in the femoral head (left) and the acetabulum (right) (Kurrat & Oberländer, 1978).

Large portions of the femoral head are covered with articular cartilage, with higher thickness in the middle, central areas of the head (Hamill et al., 2015, p. 180) (Figure 2.2). The acetabular cartilage is horseshoe-shaped, with the facies lunata as the covered area, surrounding the uncovered central portion of the acetabulum, termed fossa acetabuli. The cartilage of the facies lunata is thickest at the upper edge (Kurrat & Oberländer, 1978) (Figure 2.2) and cartilage thickness corresponds to the pressure observed at the cartilage surface (Widmer et al., 2002). In individuals with HOA, most defects of the acetabular cartilage occur in the anterior and superior portion, and lesions in this region have been associated with greater disability compared to defects of the femoral cartilage (Kumar et al., 2013). The articular cartilage forms a functional unit with the underlying subchondral bone (Golding & Goldring, 2010; Imhof et al., 1999). Thinning of articular cartilage and thus the joint space is used to radiographically evaluate OA progression (Altman et al., 1987). However, more recent research suggests that adaptations of the subchondral bone occur prior to alterations of the articular cartilage (Figure 2.3), potentially due to the faster adaptability of bone tissue to mechanical

loads compared to cartilage tissue. Imbalance in adaptation disrupts the physiological functioning of this unit and is likely to contribute to OA development (Goldring & Goldring, 2010; Li et al., 2013).

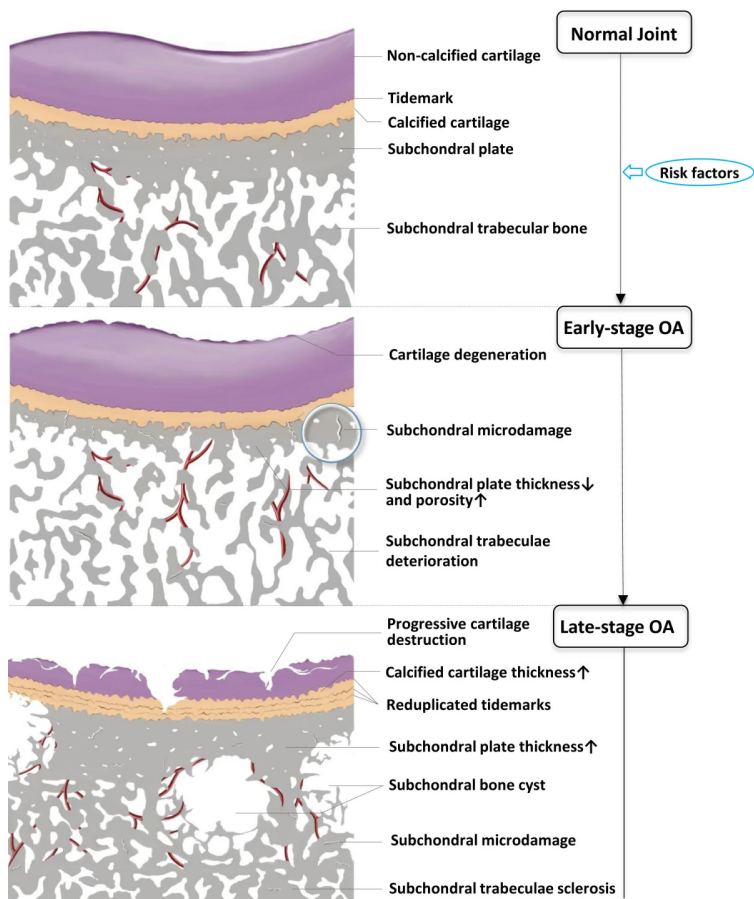


Figure 2.3: Schematic diagram of articular cartilage and subchondral bone in normal and osteoarthritis (OA) joints (Li et al., 2013).

The hip joint is moved and stabilised by 21 mono- and biarticular muscles (Neumann, 2010). While excessive hip extension is restricted, especially by the ligamentum iliofemorale, hip flexion is not constrained by ligaments, and the muscles responsible for hip extension are the strongest (Wurzinger, 2007, p. 334). Of these, the M. gluteus maximus is the strongest extensor of the hip, while the strongest flexor is the M. iliopsoas, assisted by the Mm. sartorius, rectus femoris and tensor fasciae latae. The most important hip abductors are the Mm. glu-

teus medius (GMed) and minimus, while important adductors are the Mm. adductor longus, brevis and magnus as well as the M. gracilis (Wurzinger, 2007, p. 342). Most hip muscles can contribute to movement in multiple planes, and muscle function also varies depending on the joint position (Anderson & Blake, 1994; Delp et al., 1999) (Figure 2.4). In individuals with HOA decreased muscle strength has been found in hip flexor (M. Arokoski et al., 2002; Judd et al., 2014; Loureiro et al., 2013), hip extensor (Judd et al., 2014; Loureiro et al., 2013), hip adductor (M. Arokoski et al., 2002) and hip abductor muscles (M. Arokoski et al., 2002; Burgess et al., 2022; Judd et al., 2014; Loureiro et al., 2013; Marshall et al., 2016) and also in knee extensor and knee flexor muscles (Burgess et al., 2022; Judd et al., 2014; Loureiro et al., 2013; Rydevik et al., 2010). This reduction in strength might be due to a loss in muscle volume and reduced muscle quality (Grimaldi et al., 2009; Loureiro et al., 2013; Rasch et al., 2007), but results on the underlying cause are conflicting (Marshall et al., 2016).

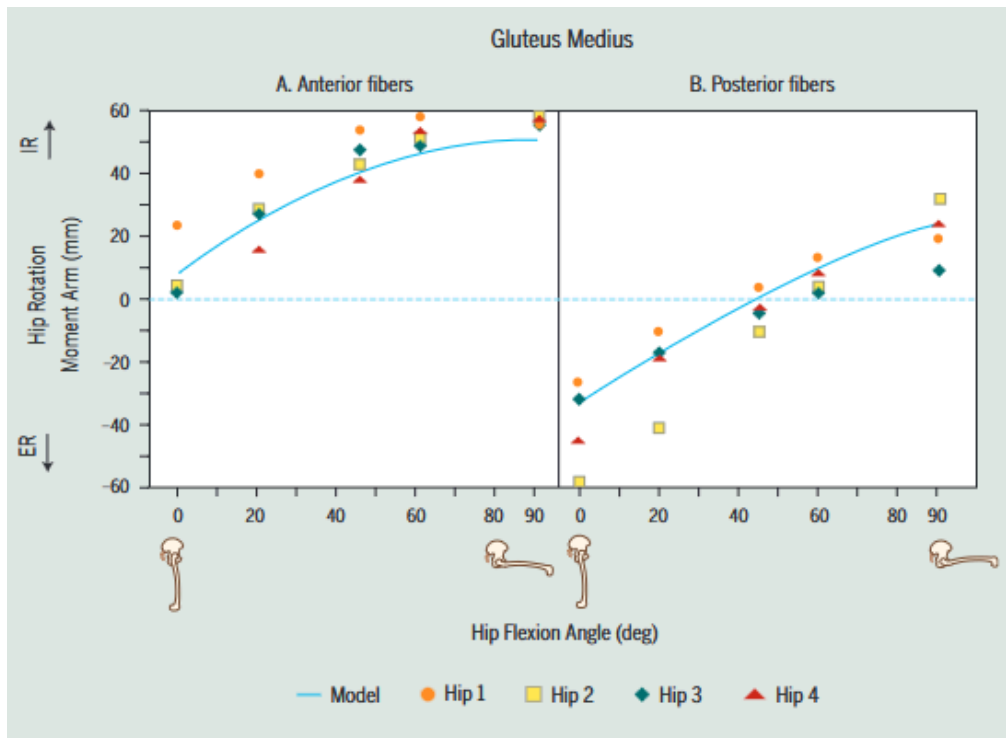


Figure 2.4: Horizontal plane rotational moment arms for two sets of fibres of the gluteus medius, plotted as a function of hip flexion.

Note. IR = internal rotation; ER = external rotation. Data: Delp et al. (1999); Figure: Neumann (2010).

In summary, the hip joint possesses a unique anatomy that provides high stability while simultaneously allowing for mobility. This anatomy diverges drastically from those of other joints frequently affected by OA, as the knee or finger joints, which stresses the need for therapy options tailored to the demands of the hip joint. Moreover, alterations in hip morphology and function appear to play a crucial role in the initial development of HOA and also the progression of this disease.

2.4 Pathomechanism and risk factors for hip osteoarthritis

The pathomechanism of HOA is thought to be “[...] initiated by micro- and macro-injury, activating maladaptive repair responses [...] which manifests first as a molecular derangement (abnormal joint tissue metabolism) followed by anatomic, and/or physiologic derangements (characterised by cartilage degradation, bone remodelling, osteophyte formation, joint inflammation and loss of normal joint function)” (Kraus et al., 2015).

Based on the initiating cause, classically HOA is classified as primary or secondary HOA, with primary HOA having an idiopathic origin (no known cause) and secondary HOA developing as a result of a defined disorder or trauma, such as congenital hip dislocation, femoroacetabular impingement (FAI) or Perthes disease (Lespasio et al., 2018; Sims, 1999). However, there is increasing evidence that most cases classified as primary HOA are actually secondary HOA due to subtle changes in hip morphology or metabolic abnormalities (Ganz et al., 2008; Harris, 1986; Jenei-Lanzl & Zaucke, 2025).

Several risk factors for the development of HOA disease or illness have been identified. These risk factors can be categorised into those at the individual level and those at the joint level. Some of these risk factors, such as age, sex, genetics, or ethnicity at the individual level, and joint morphology at the joint level, cannot be modified (N. Murphy et al., 2016). In contrast to KOA, it has been shown that the influence of genetics is greater in HOA (Magnusson et al., 2017), and that female sex is not associated with a higher risk of developing HOA (Fan et al., 2023; Pereira et al., 2011; Srikanth et al., 2005).

Other identified risk factors are modifiable and, therefore, can be targeted for the prevention of HOA, such as weight, occupation and muscular function. Thereby, a higher body mass index (BMI) is associated with an increased risk of developing HOA (Jiang et al., 2011). However, it remains unclear whether this association is caused by weight-induced higher hip load-

ing (Sanford et al., 2014) or metabolic changes due to obesity (Forsythe et al., 2008). Furthermore, occupations involving prolonged standing and heavy lifting (Croft et al., 1992; Lievense et al., 2001), as well as long-term participation in high-impact sports (L'Hermette et al., 2006; Vigdorchik et al., 2017) have been linked to increased risk of HOA. Finally, muscle weakness in general (Hurley, 1999) and specifically of the knee extensor muscles has been identified as a risk factor for developing HOA (Hootman et al., 2004).

2.5 Non-pharmacological conservative treatment of hip osteoarthritis

As the development of proactive strategies for HOA prevention is challenging, current treatment is primarily reactive, starting at the onset of illness and focusing on decelerating, or ideally halting, disease progression, reducing pain, and increasing functional capacity (Perrot, 2012; Runhaar & Bierma-Zeinstra, 2022). In doing so, treatment of OA is often structured as a pyramid with education, weight loss or exercise as first-line treatment, pharmacological interventions, medical aids or therapist-delivered interventions as second-line treatments and surgical interventions such as THR as third-line interventions (Dieppe & Lohmander, 2005; Roos & Juhl, 2012) (Figure 2.5). In the past, much research on treatment for HOA has concentrated on end-stage HOA. By shifting the management focus to the earlier stages of the disease, it may be possible to slow disease progression, enhance patients' quality of life, and ultimately reduce the need for hip replacement surgeries (N. Murphy et al., 2016). The American College of Rheumatology strongly recommends weight loss for overweight or obese patients, self-efficacy and self-management programs, exercise, tai-chi, and cane use as conservative non-pharmacological treatment options (Kolasinski et al., 2020). Similar recommendations are made by the Osteoarthritis Research Society International, including structured land-based exercise programs, patient education and self-management programs, use of gait aids, mind-body exercises (such as tai-chi and yoga), but not dietary weight management (Bannuru et al., 2019).

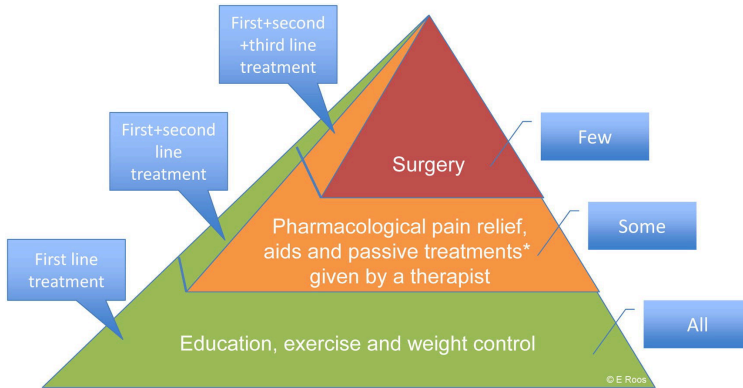


Figure 2.5: The osteoarthritis treatment pyramid (Roos & Juhl, 2012).

While weight loss is limited to overweight patients, cane use is only recommended for patients with a sufficiently large impact of HOA on locomotion (Kolasinski et al., 2020). Therefore, for an active population experiencing mild-to-moderate symptoms, exercise, tai chi, and yoga, along with patient education, self-management, and self-efficacy programs, remain as non-pharmacological conservative treatment options. While exercise has been shown to benefit pain and function (Teirlinck et al., 2023), participation in exercise is likely to depend on the presence of pain (Rosemann et al., 2008) and the level of physical function (Veenhof et al., 2012).

Another conservative non-pharmacological treatment is bracing. While for individuals with KOA, brace use has been extensively studied (Beaudreuil et al., 2009; Cudejko et al., 2018; Duivenvoorden et al., 2015) and is recommended for conservative KOA treatment (Kolasinski et al., 2020), only a limited number of studies exist analysing their therapeutic effectiveness in individuals with HOA. Consequently, bracing is not included in treatment recommendations for HOA because of insufficient research.

The few studies on hip bracing in HOA have shown improvements in functional tests (E. Sato et al., 2012) and hip pain (T. Sato et al., 2008). Thus, brace use might be an option to alleviate pain and improve function in individuals with HOA, consequently enhancing participation in daily activities and exercise. Currently, hip braces are, if at all, used after hip surgery to restrict excessive joint movement (Yonclas et al., 2006). Besides the limited availability of studies verifying the effectiveness of bracing for conservative HOA treatment, discomfort caused by the weight and stiffness of hip braces designed to mechanically stabilise or unload the hip joint may hinder their acceptability (Eyles et al., 2022; Kemker et al., 2021; Newcomb

et al., 2018). Furthermore, rigid braces that restrict motion may not be appropriate for individuals with mild or moderate symptoms (Shiba et al., 1998).

Elastic knee braces can reduce pain and improve self-reported physical function in individuals with KOA (Cudejko et al., 2018). They have also positively influenced neuromuscular control in healthy individuals (Bodendorfer et al., 2019) and proprioception in healthy individuals (Baltaci et al., 2011), individuals with KOA (Barrett et al., 1991; Birmingham et al., 2001) and individuals without an anterior cruciate ligament (Birmingham et al., 2001a). Thereby, increased skin temperature and mild compression are thought to enhance proprioception, leading to improved stability and reduced joint load, finally resulting in increased physical function and reduced pain (Cudejko et al., 2018) (Figure 2.6). However, to date, this approach has not been evaluated for the treatment of HOA.

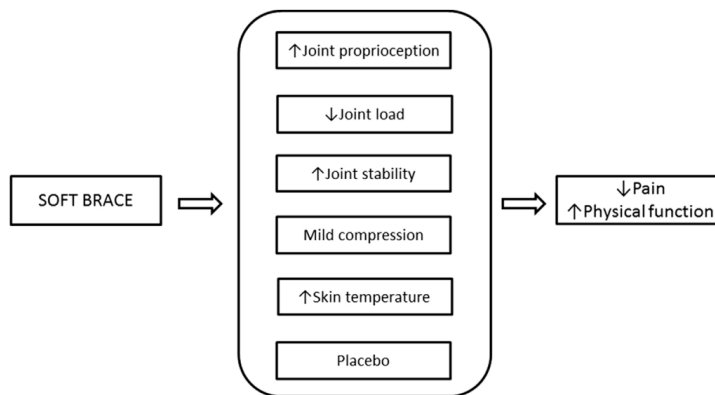


Figure 2.6: Suggested mechanisms explaining the effects of soft braces on pain and physical function in knee osteoarthritis (Cudejko et al., 2018).

2.6 Locomotion biomechanics, muscle activity, physical function and proprioception in individuals with hip osteoarthritis

Research on HOA symptoms, function and effectiveness of treatment is often based on patient-reported outcomes obtained using questionnaires (Nilsdotter & Bremler, 2011; Terwee et al., 2011; Weick et al., 2020). The most frequently used questionnaires in individuals with HOA are the Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC), the Hip Osteoarthritis Outcome Score (HOOS), the Oxford Hip Score (OHS) and

the Harris Hip Score (HHS) (Gagnier et al., 2018). Data from these questionnaires are often combined with functional tests such as the 40 m Fast Paced Walk Test, the Thirty Second Chair Stand, the Timed Up and Go (TUG), the Stair Climb Test or the 6-Minute-Walking-Test (6MWT) (A. Davis et al., 2022; Dobson et al., 2013) and individuals with HOA show lower performance during these tests (M. Arokoski et al., 2004; Judd et al., 2014; Rydevik et al., 2010).

While questionnaires and functional tests are easy to use, especially in clinical settings, they provide only limited insights into the underlying mechanisms of functional limitations (Ornetti et al., 2010). A more in-depth analysis is possible with biomechanical motion analysis, which focuses on collecting objective, quantitative data regarding the mechanics of the musculoskeletal system while performing a motor task (Cappozzo et al., 2005; Winter, 2009). Objective outcomes such as kinematics, including spatiotemporal parameters and joint angles, kinetics, incorporating joint moments and joint contact forces or measures of muscle activity can be obtained (Ornetti et al., 2010; Winter, 2009). Analyses of these parameters enhance our understanding of injury prevention, effective movement and treatment (Wallard et al., 2017).

Motion analysis of individuals with HOA has been mostly conducted during level walking. In this context, differences in spatiotemporal parameters, such as reduced walking speed, step length and stride length, have been described for all severity levels of HOA compared to individuals without HOA (Constantinou et al., 2014).

The most prominent kinematic feature of HOA gait is a reduced sagittal plane hip RoM (Eitzen et al., 2012; Nüesch et al., 2021) caused by a decreased peak extension angle (Altman et al., 1991; Constantinou et al., 2017; Eitzen et al., 2012; Hurwitz et al., 1997; Kumar et al., 2015; Pua et al., 2009; Watelain et al., 2001) and accompanied by a discontinuity of the hip extension movement during the late stance phase (ST) (Foucher et al., 2012; Hurwitz et al., 1997). These deviations have been found in individuals with mild-to-moderate (Constantinou et al., 2017; Leigh et al., 2016; Rutherford et al., 2015; Watelain et al., 2001) and severe HOA (Hurwitz et al., 1997; Meyer et al., 2018; Tateuchi et al., 2014) and increase with HOA severity (Rutherford et al., 2015).

Frontal and transverse plane hip kinematics are less frequently analysed, but lower frontal and transverse hip RoM in individuals with moderate or severe HOA (Rutherford et al., 2015; Zügner et al., 2018) were found, mainly caused by lower hip adduction (Leigh et al., 2016; Meyer et al., 2018; Tateuchi et al., 2014), abduction (Kubota et al., 2007; Rutherford et al., 2015) and internal rotation angles (Leigh et al., 2016; Rutherford et al., 2015).

Hip osteoarthritis does not exclusively affect hip kinematics. Changes in pelvic kinematics have been found and interpreted as compensatory mechanisms for limited hip mobility

(Murray et al., 1971; Watelain et al., 2001). Thereby, increased sagittal RoM (Watelain et al., 2001) due to a more pronounced anterior tilt (Kubota et al., 2007; van Drongelen et al., 2021) as well as lower frontal pelvis RoM have been described (Bolink et al., 2015; Constantinou et al., 2017; Reininga et al., 2012). Likewise, kinematics of the contralateral hip joint (Ornetti et al., 2011), the ipsi- and contralateral knee joints (Aminian et al., 2004; Foucher & Wimmer, 2012; Porta et al., 2021; Stief et al., 2018) and the ankle joints (Ismailidis et al., 2020) are also affected by HOA.

By integrating joint kinematics and ground reaction force (GRF) information, joint moments can be calculated using an inverse dynamics approach (Nigg & Herzog, 2007). The obtained joint moments are widely used as a surrogate measure for estimating joint loading (Diamond et al., 2018). Lower external hip moments in all directions have been observed in individuals with HOA during level walking (Eitzen et al., 2012; Foucher, 2017; Hurwitz et al., 1997; Kubota et al., 2007; Meyer et al., 2018; Tateuchi et al., 2014; Watelain et al., 2001). A meta-analysis revealed that lower joint moments cannot be explained solely by slower walking speed, but also stem from altered movement patterns. Furthermore, the reduction in joint moments appears to be related to the severity of HOA, with individuals in the end stage of HOA demonstrating lower joint moments, unlike those with moderate HOA (Diamond et al., 2018).

The growing use of complex musculoskeletal models has enhanced the analysis and consequently, the understanding of joint contact forces in individuals with HOA and their interaction with joint moments and muscular activity. By that means, individuals with HOA have been found to have lower hip and knee contact forces (Diamond et al., 2020; van Rossom et al., 2023). Relating external hip moments during level walking to the occurring hip contact forces showed that the first peak of hip contact force is associated with the hip adduction moment, while the second peak of hip contact force is associated with the adduction and flexion moments (Wesseling et al., 2015). The hip contact forces during level walking are predominantly caused by hip muscle activity rather than gravity; especially the GMed and to a lesser extent the gluteus maximus contribute to anterior, superior and medial peak contact forces of 4.3 times body weight (Correa et al., 2010). Lower hip extensor and hip flexor muscle forces likely cause the observed reduced hip loading (Diamond et al., 2024), in spite of an increase in muscular co-contraction of hip flexors and extensors as well as hip adductors and abductors (Diamond et al., 2020).

Despite the large number of observed modifications in locomotion biomechanics of individuals with HOA, it remains unclear whether these modifications delay or drive the degeneration of the affected joint (Eitzen et al., 2012; Hurwitz et al., 1997; Leigh et al., 2016). However, persistent gait modifications likely lead to muscle weakness in less-used muscles (Hurwitz et

al., 1997; Meyer et al., 2018). Analyses of muscle activity during level walking have shown increased GMed muscle activity in individuals with end-stage HOA during level walking and up/down stepping (Dwyer et al., 2013; Sims, 2002), increased quadriceps activity at mid-stance during level walking (Rutherford et al., 2015b), and increased gluteus minimus activity during level walking (Zacharias et al., 2019). These increases in muscle activity may compensate for reduced muscle strength (Rutherford et al., 2015a), yet this remains speculative.

Besides the biomechanical analysis of locomotion tasks, postural control is often assessed due to its high relevance in fall prevention (Granacher et al., 2008; Rubenstein, 2006). Postural control encompasses all actions involved in maintaining, achieving or restoring a state of balance, which means that the line of gravity falls within the base of support (BoS) (Pollock et al., 2000). Postural control is a complex function that involves sensory input, central processing, and the neuromuscular response (Sziver et al., 2016). The sensory input comprises, besides vestibular and visual information, afferent internal information from sensory receptors located in our skin, joints, tendons, and muscles (Gardner, 2021; Proske & Gandevia, 2012), termed proprioception. Individuals with HOA have a significantly reduced number of these receptors (Moraes et al., 2011), and a loss of vibratory sense has been described in individuals with HOA, leading to a decrease in joint position sense of the hip (Alkhamis et al., 2024) and the ipsilateral knee (Shakoor et al., 2014). Moreover, proprioceptive inaccuracy has been described as a risk factor for functional decline in individuals with HOA (Dekker et al., 2009).

However, the impact of altered muscle force, muscle activity and joint proprioception on postural control remains unclear, as the results vary strongly. Some studies showed decreased postural control compared to the contralateral limb or healthy controls in static (Sziver et al., 2016; Truszczyńska et al., 2016) and dynamic balance tasks (Kiss, 2010; Majewski et al., 2005), while others did not (Alkhamis et al., 2024; J. Arokoski et al., 2006; Rasch et al., 2010). One explanation for the heterogeneity of results is that postural control is task-specific and thus, results from different tasks are likely not comparable (Giboin et al., 2015; Ringhof & Stein, 2018). This emphasises the importance of evaluating postural control in relevant everyday movements, such as stair walking, which has often been linked to falls (Jacobs, 2016).

2.7 Biomechanical effect of hip bracing for hip osteoarthritis

As previously described (chapter 2.5), hip bracing was able to decrease pain and increase function in individuals with HOA (E. Sato et al., 2012; T. Sato et al., 2008; Yamaji et al., 2009). However, the underlying mechanisms are rarely determined. One study evaluating the Hip Joint Moment Reduction Brace in a small cohort of healthy individuals found that bracing immediately reduced hip abductor muscle activity during walking by 32.6 % (Shiba et al., 1998). An evaluation of the WISH brace, constructed to reduce hip adduction and exert pressure to the greater trochanter, showed an increase in the first peak of the vertical GRF (Yamaji et al., 2009), but the sample size was small, and no joint kinematics or kinetics were reported, limiting the insights of these results. Another brace for individuals with HOA aimed to reduce the hip adduction angle and moment as well as the hip internal rotation angle by externally applying hip abduction and external rotation forces. The intended effects were achieved during level walking. However, the effect on pain perception varied strongly between participants (Nérot & Nicholls, 2017).

With regard to the numerous observed changes in gait patterns, the braces previously described mainly reinforced mechanisms already present in individuals with HOA, such as reduced hip adduction angle and moment or alteration of hip abductor muscle activity. However, it has been postulated that treatment for HOA should aim at pain relief and normalisation of the gait pattern to avoid secondary damage to adjoining soft tissue areas and joints (Hurwitz et al., 1997; van Drongelen et al., 2020). Moreover, rigid brace concepts as evaluated in prior studies might, despite their positive effects on pain and function, possess limited applicability in individuals with mild or moderate symptoms due to the disadvantages of low comfort and restricted wearability during more dynamic movements.

Lastly, a comprehensive analysis of bracing effects, including function, pain, postural control and biomechanics as well as their interaction, is still lacking.

3 Aim of this thesis

In summary, it can be said that HOA is a highly prevalent disease that affects hip joint function and thus impacts locomotion. However, much of the previous research has focused on individuals with severe HOA and its effect on straight walking biomechanics. Consequently, knowledge of hip function in individuals in the early stages of the disease and its impact on other locomotion tasks is limited. Similarly, research in the field of HOA treatment has often focused on surgical joint replacement for end-stage HOA. Conservative treatment options for earlier disease stages are less frequently studied, and treatment recommendations are often extrapolated from results on KOA, despite major differences in joint anatomy. In this context, rigid hip braces, designed following the example of unloading knee braces, have been evaluated for HOA treatment. While these braces have been shown to have a positive effect on pain and function, they are not applicable in individuals with mild or moderate HOA due to restricted wearability during more dynamic movements. Lastly, a comprehensive analysis of the effects of bracing and the underlying mechanisms is lacking.

The aim of this thesis is therefore to comprehensively assess the effects of

1. mild-to-moderate HOA on functional capacity, hip proprioception, postural control and locomotion biomechanics.
2. hip bracing with an elastic hip brace as conservative therapy for mild-to-moderate HOA.

To realise this aim, two original research studies and one systematic literature review and meta-analysis were conducted.

To obtain a comprehensive analysis of kinematic changes observed during locomotion tasks in individuals with HOA, a systematic literature review was conducted, and the results of individual studies were pooled by means of a meta-analysis (chapter 4).

Manuscript I – Effects of Hip Osteoarthritis on Lower Body Joint Kinematics During Locomotion Tasks: A Systematic Review and Meta-Analysis

In the first research study, a measurement method for assessing the joint position sense of the hip joint was refined, and inter-session reliability of this method was determined. The results of this study are presented in chapter 5.

Manuscript II – Inter-session Reliability of Sagittal Plane Hip Proprioception Measurement in Healthy Young Adults

The largest part of this thesis is a combined case-control and intervention study examining the effects of mild-to-moderate HOA as well as short- and medium-term hip bracing on locomotion biomechanics, proprioception, pain, and function. This study includes a biomechanical movement analysis of level walking, turning and stair walking. Chapters 6 to 10 contain separate manuscripts that are based on the dataset acquired during this study and focus on different aspects of the study's aim.

A biomechanical analysis of turning movements was included in the study as turning requires substantial motion in the transverse plane. However, as knowledge on the biomechanics of turning in comparison to level walking is sparse, in chapter 6, the required hip mobility and observed hip moments during these tasks were compared and differences between healthy individuals and individuals with HOA were assessed.

Manuscript III – Mobility Requirements and Joint Loading during Straight Walking and 90° Turns in Healthy Older People and Those with Hip Osteoarthritis

The biomechanical analysis of stair walking was included in the study as it is a demanding task with high joint loading, and it is relevant to the prevention of falls. Chapter 7 presents the analysis of hip and trunk biomechanics and dynamic postural control during steady-state stair ascent and stair descent in individuals with and without HOA.

Manuscript IV – Hip and Trunk Biomechanics and Dynamic Balance During Steady-State Stair Walking in People With Mild-to-Moderate Hip Osteoarthritis

Chapter 8, firstly, contains the analysis of the effect of mild-to-moderate HOA on functional capacity, hip proprioception and level walking biomechanics, and secondly, the assessment of short- and medium-term bracing effects on the hip and pelvis biomechanics during level walking, as one of the most fundamental human movements.

Manuscript V – Effects of Hip Bracing on Gait Biomechanics, Pain and Function in Individuals With Mild to Moderate Hip Osteoarthritis

Chapter 9 contains the analysis of the effects of mild-to-moderate HOA and short- and medium-term bracing on turning biomechanics.

Manuscript VI – Mild to Moderate Hip Osteoarthritis and Hip Bracing Influence Hip and Knee Biomechanics During 90° Turns While Walking

Chapter 10 includes an analysis of medium-term bracing effects on hip, pelvis and trunk biomechanics during stair ascent and descent and relates observed changes in pain perception with brace-induced changes in stair walking biomechanics.

Manuscript VII – Effect of Hip Bracing on Stair Walking Biomechanics and Pain in Patients With Mild-to-Moderate Hip Osteoarthritis: An Intervention Study

Chapter 11 summarises the key findings from the aforementioned studies, offers a comprehensive discussion of the results in relation to existing literature, presents implications and recommendations for future research, and addresses the limitations of the current research study.

Finally, in chapter 12, a short conclusion of this thesis is drawn.

4 Manuscript I - Effects of hip osteoarthritis on lower body joint kinematics during locomotion tasks: a systematic review and meta-analysis

Slightly modified version of the manuscript published as:

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4.1 Abstract

Motion analysis can be used to gain information needed for disease diagnosis as well as for the design and evaluation of intervention strategies in patients with HOA. Thereby, joint kinematics might be of great interest due to their discriminative capacity and accessibility, especially with regard to the growing usage of wearable sensors for motion analysis. So far, no comprehensive literature review on lower limb joint kinematics of patients with HOA exists. Thus, the aim of this systematic review and meta-analysis was to synthesise existing literature on lower body joint kinematics of persons with HOA compared to those of healthy controls during locomotion tasks.

Three databases were searched for studies on pelvis, hip, knee and ankle kinematics in individuals with HOA compared to healthy controls during locomotion tasks. Standardised mean differences were calculated and pooled using a random-effects model. Where possible, subgroup analyses were conducted. Risk of bias was assessed with the Downs and Black checklist.

A total of 47 reports from 35 individual studies were included in this review. Most studies analysed walking, and only a few studies analysed stair walking or turning while walking. Most group differences were found in ipsi- and contralateral three-dimensional hip and sagit-

tal knee angles, with reduced ranges of motion in HOA participants. Differences between individuals with mild-to-moderate and severe HOA were found, with larger effects in individuals with severe HOA. Additionally, stair walking and turning while walking might be promising extensions in clinical gait analysis due to their elevated requirements for joint mobility. Large between-study heterogeneity was observed, and future studies have to clarify the effects of OA severity, laterality, age, gender, study design and movement execution on lower limb joint kinematics.

4.2 Introduction

Hip osteoarthritis is a common joint disease with high prevalence, especially in the elderly (Blanco et al., 2021). Due to pain and limited function, HOA has a strong impact on the quality of life of the affected persons (Salaffi et al., 2005).

Data from instrumented movement analysis can be helpful for disease diagnosis as well as for the design and evaluation of patient-specific intervention strategies in HOA populations (Broström et al., 2012; Ornetti et al., 2010). Depending on the data required, motion analysis can be obtained at different levels of complexity. While spatiotemporal gait characteristics are relatively easy to record (R. Davis, 1997), they might contain only limited informative value. Analysis of joint dynamics, first, requires simultaneous recording of GRF data and, secondly, advanced methods of modelling (Camomilla et al., 2017). Due to the complexity and large time requirements, the applicability in the clinical setting might be limited (Rahman et al., 2015). Diamond et al. (2018) reviewed the existing literature on external hip flexion and adduction moments in individuals with HOA. They found a reduction of both moments in patients with severe HOA but not in those with mild-to-moderate symptoms. Hence, adjustments of joint loading might take place at a later disease stage and thus might not be suitable for the early diagnosis of HOA. Similarly, Emmerzaal and colleagues (Emmerzaal et al., 2022) found lower or only slightly higher classification results between HOA and healthy individuals when joint dynamics combined with joint kinematics were used as input variables compared to kinematic variables only.

Consequently, analysis of joint kinematics might be the right balance between information acquisition and feasibility, especially in the clinical context. Thereby, wearable sensors' increasing dissemination and popularity for the analysis of joint kinematics also have to be considered. With the availability of easy-to-use sensors and robust software tools for the estimation of joint angles, widespread usage of movement analysis in the clinical setting

becomes feasible (Petraglia et al., 2019). In the past, several studies have shown kinematic changes at the hip joint in persons with varying degrees of HOA (Diamond et al., 2020; Eitzen et al., 2012; Kumar et al., 2015). Additionally, previous studies have shown that changes in gait patterns not only concern the hip joint itself but also the knee (Rutherford et al., 2015b), ankle (Leigh et al., 2016), pelvis (Watelain et al., 2001) and upper body kinematics (Reininga et al., 2012). Therefore, kinematics in general are of great interest for the characterisation of gait changes associated with HOA.

Gait, as one of the most fundamental movements of daily life, has often been used in the analysis of disease effects on movement kinematics, and previous reviews (Constantinou et al., 2014; Diamond et al., 2018; Ornetti et al., 2010) focused on gait deviations caused by HOA. However, other locomotion tasks, such as stair ascent and descent, might impose higher demands regarding the required RoM (Grimmer et al., 2020), and thus their analysis might add essential insights into the movement restrictions and adaptations of people with HOA. A recently published article has shown that the most accurate classification between individuals with and without HOA was achieved using joint angle data from stair walking trials compared to walking or stationary tasks such as lunges, etc. (Emmerzaal et al., 2022).

Biomechanical gait analysis studies are often time-consuming (Broström et al., 2012; Ornetti et al., 2010) and therefore only include limited sample sizes (Diamond et al., 2018). Additionally, individuals with HOA show heterogeneous disease characteristics such as varying degrees of functional and radiographic disease severity, uni- or bilateral involvement, as well as primary or secondary HOA cause (Diamond et al., 2018). An aggregation of multiple studies and a collective evaluation of the results, as well as the evaluation of specific sub-groups, might therefore add essential insights on the impact of HOA on movement biomechanics.

However, to the best of our knowledge, no comprehensive review exists regarding the kinematic changes observed during locomotion tasks in the presence of HOA.

Therefore, the objective of the present systematic review was to summarise the current state of research on lower-limb joint kinematics during locomotion tasks, such as gait or stair walking, in individuals diagnosed with HOA compared to healthy controls. Where possible, a con-joint analysis of previous results in terms of a meta-analysis was performed, allowing special attention to be paid to the influence of HOA severity and uni- or bilateral involvement.

4.3 Materials and Methods

This review was conducted according to the Preferred Reporting Items for Systematic Review and Meta-Analysis Statement (PRISMA) (Page et al., 2021) and registered in the International Prospective Register of Systematic Reviews (PROSPERO, no. CRD42021238237).

4.3.1 Information sources, search strategy and screening process

Eligible reports were searched in three electronic databases (PubMed (MEDLINE, PubMed Central, and additional PubMed records), Web of Science and Scopus) on August 2nd 2022.

The title, abstract and keywords were screened for disease description (Coxarthr*, “degenerative joint” AND hip, hip AND osteoarthr*), the outcome parameter (kinematic*, angle*, “range of motion”, mobility, pattern, goniometric*, biomechanic*) and the movement task. (gait, walk*, locomotion, ambulat*, stair*, movement). Titles including fracture, perthes, amputee, rheumat*, arthroscopy or arthroplasty were excluded. The detailed search terms for each database can be found in the Appendix (Table A.1).

All records from the databases were imported into a reference manager (Citavi 6, Swiss Academic Software GmbH, Switzerland) and duplicates were removed. The reference lists of included articles were screened manually for additional eligible studies.

The literature search, title and abstract screening as well as full text analysis for eligibility were performed individually by two researchers (HS & SSp). Discrepancies were resolved by discussion between the two authors, and if no consensus was reached, a third researcher (TS) was consulted.

4.3.2 Eligibility criteria

Original research studies written in English, evaluating pelvis or lower extremity (hip, knee, ankle) joint kinematics during locomotion tasks (e.g. level walking, running or stair walking) in a cohort of individuals with HOA were eligible for this review. No restrictions were made regarding the OA severity, unilateral or bilateral involvement or whether the HOA was primary or secondary. Data from the HOA group had to be compared to a healthy control group (CON). Accepted outcome parameters were a quantitative description of lower body joint

angle parameters (mean/median with a measure of dispersion) and/or the presence of p-values from the comparison of lower body joint angle parameters. Studies with an intervention were included using the pre-intervention data if applicable. A detailed description of the eligibility criteria can be found in Table 4.1.

Table 4.1: Inclusion and exclusion criteria following a modified PECO scheme.

	Inclusion criteria	Exclusion criteria
Population	Humans with hip osteoarthritis (uni- & bilateral)	Individuals with other diseases (e.g. hip dysplasia, knee osteoarthritis) or endoprosthesis
Exposure	Analysis of locomotion tasks (e.g. gait, stair walking)	Stationary movements (e.g. sit to stand, one-leg stand), use of walking aids or handrails
Comparator	Healthy control group	No control group or contralateral limb as control
Outcome	Report of kinematic data on lower body joint angles (hip, knee, ankle) or pelvis movement	Sole report of temporal-spatial-parameters or other kinematic data (e.g. toe-out angles), or analyses of kinematic coordination (whole body analysis, coupling angles, asymmetry measures)
Language	English language	Other languages
Format	Original full text paper	Reviews, conference proceedings, case studies, etc.

4.3.3 Data collection process

One researcher (HS) extracted the data from the retrieved reports using a predefined spreadsheet including the following sections: study design, number of HOA and CON participants, participant characteristics (i.e. age, gender, weight, height, BMI, radiographic disease severity, functional disease severity, uni-/bilateral involvement, primary/secondary OA), analysed movements (e.g. walking, stair climbing) and testing conditions (treadmill or overground, number of stairs, prescribed or self-selected movement velocity), measurement system and assessed joints (e.g. hip sagittal plane). Extracted data were randomly cross-checked by a second researcher (SSp).

The extracted data were analysed for indications of multiple reports from the same study. In case of doubt, report authors were contacted for clarification. Extracted data of the kinematic variables were summarised in tables and grouped according to planes of motion and joint location.

Disease severity was categorised as mild, moderate or severe based on the information in the reports. End-stage HOA as well as patients scheduled for THR were classified as severe.

4.3.4 Synthesis methods

Peak angles were converted if necessary to obtain a unified definition of the angle direction. If both limbs were reported for the CON participants, data from the right limb were used. One study (Zügner et al., 2018) reported joint angles of two different examiners to calculate inter-rater reliability. For the meta-analysis, values of examiner one were extracted.

Confidence intervals for group means were transformed to standard deviations (SD) following the Cochrane handbook (Higgins et al., 2024 section 6.5.2.2).

If mean and SD were available, standardised mean differences (SMD) and 95 % confidence intervals (95%CI) were calculated for all variables by dividing the difference between groups by the pooled SD (effect size Cohen's d).

Where possible (≥ 2 studies), pooled effect sizes were calculated using a random-effects model with a restricted maximum likelihood estimator and Knapp-Hartung adjustment. The random-effects model was applied to account for variability in the composition of the subject groups (e.g. HOA severity, gender, etc.) and the movement execution (e.g. movement speed). SMDs and pooled SMDs ≥ 0.2 were interpreted as a small effect, ≥ 0.5 as a moderate effect and ≥ 0.8 as a large effect (Cohen, 1988, pp. 25–27). Statistical heterogeneity was evaluated from pooled data using the I^2 statistic, with a value of 25 % considered low, 50 % considered moderate, and 75 % considered a high level of heterogeneity (Higgins et al., 2003, pp. 557–560). Additionally, prediction intervals were reported (Higgins et al., 2009; Int'Hout et al., 2016). Subgroup analyses for HOA severity and laterality (uni- / bilateral) were conducted if data from ≥ 2 studies for ≥ 2 clearly distinguishable subgroups were available. If multiple HOA subgroups from the same study were included, the sample size of the CON group was split equally among all HOA subgroups. For all analyses, the significance level was set *a priori* to $\alpha < 0.05$. The meta-analysis (including calculation of I^2 and prediction intervals) was conducted in R (version 4.2.2) using the meta package.

For several studies, multiple effect sizes were available due to multiple reports, including equal or overlapping subject samples, analyses of subgroups (e.g. men and women), or analyses of different movement conditions (e.g. walking speeds). In those cases where data of independent subgroups were presented (e.g. OA severity, gender), we recreated the summary data (weighted mean and combined SD) prior to effect size calculation (Borenstein, 2009, p. 221). If the presented data were of dependent subgroups (e.g. multiple reports for one study, multiple walking speeds analysed), one study was selected and included in

the meta-analysis based on the following criteria: (I) largest overall sample size, (II) movement condition most similar to other studies (López-López et al., 2018), (III) self-selected gait speed.

Data from studies that could not be synthesised in a meta-analysis were synthesised qualitatively.

4.3.5 Risk of bias and quality of reporting

Risk of bias and quality of reporting of all included studies were assessed individually by two researchers (HS & SSp) using the checklist created by Downs & Black (1998). Of the 27 items, eleven concerning interventions were removed (items 4, 8, 9, 13, 14, 15, 17, 19, 23, 24, 26) as previously done (Diamond et al., 2018). This resulted in a maximum score of 17, with higher scores representing a lower risk of bias. For question five, age, weight or BMI and gender were defined as principal confounders. Question 27 was answered yes if an *a priori* or *post hoc* power analysis was reported. Disagreements in initial ratings were discussed by HS and SSp to reach a consensus. If no consensus was reached, a third reviewer (TS) was consulted.

4.4 Results

The process of study identification and screening is displayed in Figure 4.1. In total, 47 reports representing 35 independent studies met the inclusion criteria and are reviewed below. Details of all included reports are listed in Table 4.2 and Table 4.3. In total, the studies included 949 participants with HOA and 886 CON participants. For 3.8 % of the HOA participants and 2.1 % of the CON participants, gender was not reported. Of the remaining participants, 56.7 % of individuals with HOA and 62.7 % of individuals without HOA were female.

Only one study included individuals with mild HOA, seven with mild-to-moderate HOA, two with moderate HOA, two studies included individuals with moderate-to-severe HOA and 21 with severe HOA. In three studies, individuals with varying degrees of HOA were included, and two studies gave no details on HOA severity.

In 20 studies, individuals with unilateral HOA were included and two studies included those with bilateral HOA. Mixed samples were included in five studies and nine studies did not give information on limb involvement. In three studies, participants were diagnosed with primary HOA. For all other studies, no information regarding primary or secondary OA was available.

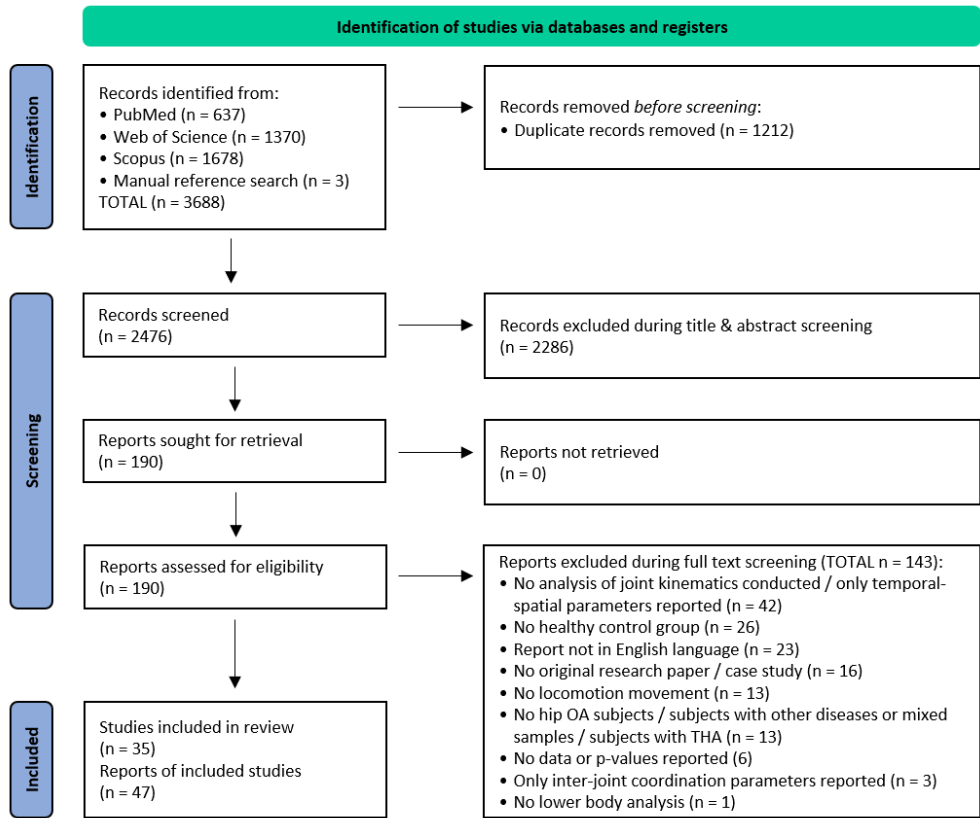


Figure 4.1: PRISMA flow diagram (Page et al., 2021).

Gait movement was analysed in 33 studies, of which 28 analysed overground walking, while five studies analysed walking on a treadmill. Two studies analysed stair ascent and descent and one study analysed turning while walking.

Hip kinematics were analysed in 31 studies, knee kinematics in 18 studies, ankle kinematics in 13 studies and pelvis kinematics in 16 studies. Thereof, ten studies (hip and knee), five studies (ankle) and two studies (pelvis) also analysed contralateral joint kinematics.

Data from 29 studies (27 studies on gait, two studies on stair walking) were included in the meta-analysis. For the sake of brevity, only forest plots of analyses with significant results are included in the text, and forest plots for all other analyses can be found in the Appendix (Figure A.1 to Figure A.21).

Table 4.2: Detailed population characteristics of the included studies.

		Population characteristics						
Study	Report	Group	n (?)	Age [years]	Height [cm]	Weight [kg]	BMI [kg/m ²]	OA characteristics (severity, laterality, cause)
A	Aminian et al. (2004)	CON	9 (-)	63 ± 4	161 ± 10	63 ± 9		Severe; unilateral
		HOA	11 (-)	60 ± 9	167 ± 5	74 ± 8		
B	Ardestani & Wimmer (2016)	CON	23 (13)	58.2 ± 9.7			28.0 ± 3.9	Mild, Moderate, Severe (Subgroups); unilateral
		HOA	45 (20)	59 ± 11				
C	Baker et al. (2016)	CON	20 (-)	63 ± 6	172 ± 9	76.0 ± 18.4	25.6 ± 5.0	Moderate, Severe (Subgroups); unilateral
		HOA	36 (-)	61.2 ± 3.8	170 ± 2.7	84.2 ± 4.1	28.9 ± 2.0	
	Rutherford et al. (2015b)	CON	20 (10)	63 ± 6	172 ± 9	76 ± 18.4	25.6 ± 5.1	Moderate; unilateral
		HOA	20 (5)	59 ± 8	170 ± 8	83 ± 16.8	28.7 ± 4.3	
	Rutherford et al. (2015a)	CON	20 (-)	62 ± 6			25.6 ± 5	Moderate, Severe (Subgroups); unilateral
		HOA	37 (-)	60.8 ± 3.4			29.3 ± 2.2	
D	Bejek et al. (2006)	CON	20 (12)	68.8 ± 9.1	169 ± 19	73.3 ± 11.4		Severe; unilateral
		HOA	20 (12)	69.7 ± 8.9	172 ± 11	70.1 ± 9.1		
E	Benedetti et al. (2012)	CON	10 (4)	59				Severe; -
		HOA	8 (3)	48.7	173	78.5		

Table 4.2: Detailed population characteristics of the included studies.

		Population characteristics						
Study	Report	Group	n (%)	Age [years]	Height [cm]	Weight [kg]	BMI [kg/m ²]	OA characteristics (severity, laterality, cause)
F	Bolink et al. (2015)	CON	20 (11)	61.0 ± 6.1	173 ± 8.4	77.2 ± 12.7	25.8 ± 3.0	Severe; unilateral
		HOA	20 (10)	63.4 ± 8.5	172 ± 9.7	81.1 ± 17.8	27.2 ± 4.9	
G	Brand & Crowninshield (1980) CON data: Crowninshield et al. (1978)	CON	15 (7)	71				
		HOA	8 (-)					
H	Constantinou et al. (2017)	CON	26 (18)	59.3 ± 7.6	169 ± 8	70.5 ± 9.3	24.8 ± 3.0	Mild/Moderate; mixed (11 bi, 16 uni)
		HOA	27 (18)	63.2 ± 7.6	166 ± 9	77.6 ± 14.2	28.0 ± 4.1	
	Diamond et al. (2020)	CON	23 (17)	60 ± 8	167 ± 8	69.9 ± 9.5	25.1 ± 3.1	Mild/Moderate; mixed (9 bi, 9 uni)
		HOA	18 (13)	65 ± 7	166 ± 10	76.2 ± 14.1	27.6 ± 4.8	
I	Eitzen et al. (2012)	CON	22 (13)	58.5 ± 8.8	172 ± 10.8	70.8 ± 15.1	23.8 ± 3.5	Mild/Moderate; mixed (10 bi, 38 uni)
		HOA	48 (29)	59.1 ± 9.5	172 ± 8.4	73.2 ± 12.2	24.6 ± 3.3	
J	Foucher (2017)	CON	159 (104)	55.7 ± 2.9			26.7 ± 2.3	Mixed; -
		HOA	150 (86)	62.3 ± 3.4			28.3 ± 2.3	
K	Foucher et al. (2007)	CON	25 (11)	57.6 ± 7.7	173 ± 9	79.2 ± 17.8		Severe; unilateral
		HOA	28 (10)	63.6 ± 7.1	172 ± 11	86.7 ± 15.8		
L	Foucher & Wimmer (2012)	CON	25 (-)	54 ± 6			28 ± 6	Severe; unilateral; primary
		HOA	26 (-)	59 ± 9			27 ± 3	
	Foucher et al. (2011)	CON	25 (13)	54 ± 6	171 ± 6.9	81.6 ± 13.3	28 ± 6	Severe; unilateral; primary
		HOA	26 (11)	60 ± 4.2	176 ± 6.3	83.9 ± 5.9	27 ± 3	
M	Hall et al. (2017)	CON	15 (11)	64.0 ± 8.7	163 ± 11	72.6 ± 11.6	23.8 ± 3.6	Mixed; mixed (9 bi, 6 uni)
		HOA	15 (8)	63.1 ± 6.7	168 ± 11	76.0 ± 11.8	28.1 ± 4.0	

Table 4.2: Detailed population characteristics of the included studies.

		Population characteristics						
Study	Report	Group	n (♀)	Age [years]	Height [cm]	Weight [kg]	BMI [kg/m ²]	OA characteristics (severity, laterality, cause)
N	Hara et al. (2016) CON data: Hara et al. (2014)	CON	6 (0)	33	173	67		Severe; -
		HOA	14 (12)	65 ± 7	154 ± 8	54 ± 9	23 ± 4	
O	Hurwitz et al. (1997)	CON	19 (7)	61 ± 8	173 ± 10	72.4 ± 15.4		Severe; unilateral
		HOA	19 (7)	60 ± 8	170 ± 10	79.2 ± 11.2		
P	Ismailidis et al. (2021)	CON	48 (30)	66.6 ± 7.2	169 ± 9	71.6 ± 12.5	25.1 ± 4.0	Severe; unilateral
		HOA	24 (10)	66.1 ± 10.3	171 ± 8	80.8 ± 13.3	27.5 ± 3.2	
	Ismailidis et al. (2020)	CON	45 (29)	66.6 ± 7.4	169 ± 8	71.0 ± 11.9	25.0 ± 4.1	Severe; unilateral
		HOA	22 (10)	66.3 ± 10.2	171 ± 8	79.7 ± 13.2	27.3 ± 3.3	
	Nüesch et al. (2021)	CON	54 (23)	66.4 ± 7.9	170 ± 9	73.1 ± 13.5	25.3 ± 4.0	Severe; unilateral
		HOA	30 (18)	64.9 ± 11.6	172 ± 8	80.6 ± 12.8	27.1 ± 3.1	
Q	Kataoka et al. (2021)	CON	15 (15)	61.2 ± 6.3	156 ± 3.7	53.5 ± 7.3		Moderate/Severe; -
		HOA	15 (15)	60.4 ± 9.6	153 ± 2.9	57.1 ± 11.4		
R	Kubota et al. (2007)	CON	12 (12)	64.3 ± 2.8	148 ± 5.1	52.5 ± 7.8	23.9 ± 2.8	Moderate/Severe; bilateral
		HOA	12 (12)	59.4 ± 11.1	151 ± 6.1	56.7 ± 7.1	24.9 ± 3.4	
S	Kumar et al. (2015)	CON	30 (16)	48.2 ± 11.4			23.3 ± 3.3	Mild/Moderate; mixed
		HOA	36 (12)	54.5 ± 8.9			24.5 ± 3.0	

Table 4.2: Detailed population characteristics of the included studies.

		Population characteristics						
Study	Report	Group	n (%)	Age [years]	Height [cm]	Weight [kg]	BMI [kg/m ²]	OA characteristics (severity, laterality, cause)
T	Leigh et al. (2016)	CON	22 (13)	53.7 ± 8.3	169 ± 9.5	76.5 ± 9.4	26.8 ± 1.5	Mild/Moderate; mixed (6 bi, 16 uni)
		HOA	22 (12)	55.9 ± 7.5	170 ± 8.0	76.8 ± 15.4	26.5 ± 4.6	
U	Meyer et al. (2018)	CON	17 (8)	52.7 ± 4.9	171 ± 10		24.1 ± 2.7	Severe; unilateral
		HOA	20 (5)	49.7 ± 9.5	173 ± 10		25.5 ± 3.2	
	Wesseling et al. (2018)	CON	18 (9)	53.0 ± 5.0	171 ± 10	69.3 ± 12.5	23.7 ± 3.1	Severe; unilateral
		HOA	14 (5)	47.3 ± 11.8	173 ± 9	76.3 ± 16.6	25.2 ± 3.7	
V	Ornetti et al. (2011)	CON	9 (7)	60.3 ± 7				Mixed; unilateral; primary
		HOA	11 (8)	60.5 ± 7			25.7 ± 6	
W	Popovic et al. (2021)	CON	30 (16)	44.7 ± 13.5			23.7 ± 2.4	Mild/Moderate; -
		HOA	42 (23)	49.6 ± 15.2			24.5 ± 3.3	
X	Porta et al. (2021)	CON	11 (5)	67.8 ± 5.4	166 ± 7.6	75.9 ± 10.1		Severe; -; primary
		HOA	11 (5)	68.3 ± 5.8	162 ± 5.4	75.4 ± 16.9		
Y	Reininga et al. (2012)	CON	30 (22)	66 ± 6	170 ± 9	69 ± 12		Severe; -
		HOA	60 (45)	59.7 ± 3.3	171 ± 2.7	77.7 ± 3.5		
Z	Schmidt et al. (2017)	CON	18 (7)	60.4 ± 8.9	173 ± 8.9	71.5 ± 13.7	23.7 ± 3.0	Severe; unilateral
		HOA	18 (7)	64.4 ± 7.4	171 ± 7.7	80.7 ± 14.4	27.5 ± 3.4	
	Stief et al. (2018)	CON	15 (6)	61.5 ± 8.0	174 ± 9	71.7 ± 14.7	23.5 ± 2.9	Severe; unilateral
		HOA	15 (6)	65.9 ± 8.6	172 ± 8	79.7 ± 9.9	27.0 ± 2.3	

Table 4.2: Detailed population characteristics of the included studies.

		Population characteristics						
Study	Report	Group	n (♀)	Age [years]	Height [cm]	Weight [kg]	BMI [kg/m ²]	OA characteristics (severity, laterality, cause)
AA	Schmitt et al. (2015)	CON	15 (7)	49.2 ± 7.1	166 ± 18	67.4 ± 11.6		Severe; unilateral
		HOA	30 (15)	54.8 ± 6.7	172 ± 10	83.2 ± 20.6		
AB	Steingrebe et al. (2022)	CON	21 (10)	63.1 ± 9.2	171 ± 8.8	74.4 ± 12.7	25.2 ± 2.7	Mild/Moderate; unilateral
		HOA	21 (10)	64.0 ± 9.6	171 ± 6.7	71.3 ± 11.9	24.2 ± 2.9	
AC	Tanaka (1998)	CON	56 (56)					- ; unilateral, bilateral (sub-groups)
		HOA	24 (24)					
AD	Tateuchi et al. (2014)	CON	13 (13)	62.6 ± 4.4	152 ± 4.9	50.6 ± 5.3		Severe; -
		HOA	14 (14)	59.3 ± 5.3	153 ± 5.5	53.3 ± 9.1		
AE	Thurston (1985)	CON	10 (-)	63.4 ± 8.1				Severe; unilateral
		HOA	20 (0)	65.1 ± 7.8				
Z & AF	van Drongelen et al. (2021)	CON	26 (16)	63.3 ± 7.9	168 ± 10	69.3 ± 12.8	24.6 ± 3.1	Severe; unilateral, bilateral (sub-groups)
		HOA	52 (32)	64.3 ± 3.1	169 ± 3.1	77.9 ± 4.0	27.1 ± 2.1	
	van Drongelen, Stetter, et al. (2021)	CON	46 (25)	64.2 ± 7.0	169 ± 10	69.0 ± 12.6	24.2 ± 2.8	Severe; unilateral
		HOA	51 (21)	60.6 ± 9.9	173 ± 7	80.3 ± 11.5	26.7 ± 2.9	
AF	van Drongelen, Kaldowski, et al. (2020)	CON	18 (7)	60.4 ± 8.0	173 ± 9	72.0 ± 13.9	23.9 ± 3.2	Severe; unilateral
		HOA	17 (8)	60.5 ± 9.9	172 ± 9	83.3 ± 15.9	28.1 ± 4.9	
	van Drongelen et al. (2019)	CON	15 (6)	61.5 ± 8.9	174 ± 9	71.7 ± 14.7	23.5 ± 2.9	Severe; unilateral
		HOA	22 (13)	62.3 ± 10.2	171 ± 10	82.4 ± 16.7	28.2 ± 4.9	

Table 4.2: Detailed population characteristics of the included studies.

		Population characteristics						
Study	Report	Group	n (%)	Age [years]	Height [cm]	Weight [kg]	BMI [kg/m ²]	OA characteristics (severity, laterality, cause)
AG	Watanabe et al. (1999)	CON	54 (54)	29.4				Mild/Moderate; unilateral
		HOA	30 (30)	31.2				
	Watanabe et al. (1998)	CON	54 (54)	29.4 ± 4.6				Mild/Moderate; unilateral
		HOA	30 (30)	31.2 ± 5.6				
AH	Watelain et al. (2001)	CON	17 (9)	63.6 ± 5.2	169 ± 6	71.1 ± 14.0		Severe; unilateral
		HOA	17 (9)	58.9 ± 7.1	167 ± 12	76.6 ± 14.7		
AI	Zügner et al. (2018)	CON	20 (10)	45.4 ± 4.4			24.3 ± 1.4	Severe; unilateral
		HOA	20 (10)	58.5 ± 5.2			28.0 ± 2.5	

Note. Data presented as mean ± SD. CON = control group; HOA = hip osteoarthritis group; OA = osteoarthritis.

4.4.1 Risk of bias and quality of reporting

The results of the risk of bias assessment are presented in Table 4.4. The mean score was 10.5 (± 2.3), with the minimum being two and the maximum being 15 (maximum achievable = 17). We did not exclude any of the studies based on their total score.

Most studies have high scores regarding reporting (Q1-3, Q5-7, Q10) and internal validity (bias, Q16, Q18, Q20). However, external validity (Q11, Q12) is hardly ever to determine as detailed information about the recruitment process is missing. Likewise, questions regarding internal validity (confounding, Q21, Q22) are often undetermined as information on CON subject recruitment and time period of subject recruitment is missing. Power analysis was only reported in six of the revised 47 reports (Q27).

Table 4.3: Details on osteoarthritis assessment method and study design of included studies.

		Osteoarthritis assessment		Study design		
Study	Report	Radio-graphic	Functional	Movement	Measurement System	Analyzed joints
A	Aminian et al. (2004)		HHS	Gait OG	IMU	Knee (I & C)
B	Ardestani & Wimmer (2016)	K-L		Gait OG	Optoelectronic	Hip (I), Knee (I), Ankle (I)
C	Baker et al. (2016)	K-L	HOOS	Gait TM	Optoelectronic	Hip (I)
	Rutherford et al. (2015b)		HOOS, WOMAC	Gait TM	Optoelectronic	Knee (I & C)
	Rutherford et al. (2015a)			Gait TM	Optoelectronic	Hip (I)
D	Bejek et al. (2006)	K-L	HHS	Gait TM	Ultrasound	Hip (I & C), Knee (I & C), Pelvis
E	Benedetti et al. (2012)			Gait OG	Optoelectronic	Hip (I), Pelvis
F	Bolink et al. (2015)	K-L		Gait OG	IMU	Pelvis
G	Brand & Crowninshield (1980) CON data: Crowninshield et al. (1978)			Gait OG	Biplanar photography	Hip (I)
H	Constantinou et al. (2017)	K-L	HHS	Gait OG	Optoelectronic	Hip (I), Pelvis
	Diamond et al. (2020)	K-L	HHS	Gait OG	Optoelectronic	Hip (I)
I	Eitzen et al. (2012)			Gait OG	Optoelectronic	Hip (I), Knee (I), Ankle (I)
J	Foucher (2017)	K-L		Gait OG	Optoelectronic	Hip (I)
K	Foucher et al. (2007)		HHS	Gait OG	Optoelectronic	Hip (I)
L	Foucher & Wimmer (2012)			Gait OG	Optoelectronic	Hip (C), Knee (C)
	Foucher et al. (2011)			Gait OG	Optoelectronic	Hip (I)
M	Hall et al. (2017)	K-L	HOOS	Stairs (A & D)	Optoelectronic	Hip (I), Pelvis
N	Hara et al. (2016) CON data: Hara et al. (2014)	K-L		Gait TM	Continuous radiographic imaging	Hip (I), Pelvis
O	Hurwitz et al. (1997)		HHS	Gait OG	Optoelectronic	Hip (I & C)

Table 4.3: Details on osteoarthritis assessment method and study design of included studies.

		Osteoarthritis assessment		Study design		
Study	Report	Radio-graphic	Functional	Movement	Measurement System	Analyzed joints
P	Ismailidis et al. (2021)	K-L	HOOS	Gait OG	IMU	Hip (I), Knee (I), Ankle (I)
	Ismailidis et al. (2020)	K-L	HOOS	Gait OG	IMU	Hip (I & C), Knee (I & C), Ankle (I & C)
	Nüesch et al. (2021)	K-L	HOOS	Gait OG	IMU	Hip (I & C), Knee (I & C), Ankle (I & C)
Q	Kataoka et al. (2021)	K-L	HHS	Gait TM	IMU	Hip (I), Knee (I), Ankle (I)
R	Kubota et al. (2007)	K-L	JOA HS	Gait OG	Optoelectronic	Hip (I), Ankle (I), Pelvis
S	Kumar et al. (2015)	K-L	HOOS	Gait OG	Optoelectronic	Hip (I)
T	Leigh et al. (2016)	K-L	WOMAC	Gait TM	Optoelectronic	Hip (I), Knee (I), Ankle (I), Pelvis
U	Meyer et al. (2018)	Tönnis		Gait OG	Optoelectronic	Hip (I)
	Wesseling et al. (2018)			Gait OG	Optoelectronic	Hip (I & C), Knee (I & C), Ankle (I & C)
V	Ornetti et al. (2011)	K-L		Gait OG	Optoelectronic	Hip (I & C), Knee (I & C), Ankle (I & C)
W	Popovic et al. (2021)	K-L	HOOS	Stairs (A & D)	Optoelectronic	Hip (I), Knee (I), Ankle (I)
X	Porta et al. (2021)	K-L		Gait OG	Optoelectronic	Hip (I & C), Knee (I & C), Ankle (I & C)
Y	Reininga et al. (2012)			Gait OG	IMU	Pelvis
Z	Schmidt et al. (2017)	K-L	HHS	Gait OG	Optoelectronic	Knee (I & C)
	Stief et al. (2018)	K-L	HHS	Gait OG	Optoelectronic	Hip (I & C), Knee (I & C), Pelvis
AA	Schmitt et al. (2015)			Gait OG	Optoelectronic	Hip (I), Knee (I), Ankle (I)
AB	Steingrebe et al. (2022)	K-L	HHS, HOOS	Gait OG	Optoelectronic	Hip (I), Pelvis
AC	Tanaka (1998)			Gait OG	Optoelectronic	Hip (I & C)
AD	Tateuchi et al. (2014)		HHS	Gait OG, Turning 45°	Optoelectronic	Hip (I), Knee (I), Ankle (I)
AE	Thurston (1985)			Gait OG	Video camera	Pelvis
Z & AF	van Drongelen, Braun, et al. (2021)	K-L	HOOS, HHS	Gait OG	Optoelectronic	Hip (I & C), Knee (I & C), Ankle (I & C), Pelvis
	van Drongelen, Stetter, et al. (2021)		HOOS, HHS	Gait OG	Optoelectronic	Hip (I), Knee (I), Pelvis

Table 4.3: Details on osteoarthritis assessment method and study design of included studies.

		Osteoarthritis assessment		Study design		
Study	Report	Radio-graphic	Functional	Movement	Measurement System	Analyzed joints
AF	van Drongelen, Kaldowski, et al. (2020)	K-L		Gait OG	Optoelectronic	Hip (I & C), Knee (I & C), Pelvis
	van Drongelen et al. (2019)			Gait OG	Optoelectronic	Hip (I & C), Knee (I & C)
AG	Watanabe et al. (1999)			Gait OG	Optoelectronic	Pelvis
	Watanabe et al. (1998)			Gait OG	Optoelectronic	Pelvis
AH	Watelain et al. (2001)	K-L	Lequesne Index	Gait OG	Optoelectronic	Hip (I), Pelvis
AI	Zügner et al. (2018)	Ahlbäck		Gait OG	Optoelectronic	Hip (I)

Note. K-L= Kellgren-Lawrence Score; HHS = Harris Hip Score; HOOS = Hip Osteoarthritis Outcome Score; WOMAC = Western Ontario and McMaster Universities Osteoarthritis Index; JOA HS = Japanese Orthopaedic Association Hip Score; OG = overground; TM = treadmill; IMU = inertial measurement unit; I = ipsilateral; C = contralateral; A = ascent; D = descent.

Table 4.4: Results of the risk of bias assessment using the checklist of Downs & Black (1998).

Study	Report	q1	q2	q3	q5	q6	q7	q10	q11	q12	q16	q18	q20	q21	q22	q25	q27	Total	Total [%]
A	Amnian et al. (2004)	1	1	0	1	1	1	0	0*	0*	1	0*	1	0*	0*	0	0	7	41
B	Ardestani & Wimmer (2016)	1	1	1	2	1	1	1	0	0*	1	0*	1	0*	1	1	0	12	71
C	Baker et al. (2016)	1	1	1	1	1	1	1	0*	0*	1	1	1	0	0*	1	0	11	65
	Rutherford et al. (2015b)	1	1	1	2	1	1	1	0*	0*	1	1	1	0	0*	1	0	12	71
	Rutherford et al. (2015a)	1	1	1	1	1	1	1	0*	0*	1	1	1	0	0*	1	0	11	65
D	Belek et al. (2006)	1	1	1	2	1	1	0	0*	0*	1	1	1	0*	0*	1	0	11	65
E	Benedetti et al. (2012)	1	1	0	1	1	1	1	0*	0*	1	1	1	0	0*	0	0	9	53
F	Bojink et al. (2015)	1	1	1	2	1	1	0	1	0*	1	1	0	0*	0*	1	0	11	65
G	Brand & Crownshield (1980) CON data: Crownshield et al. (1978)	1	1	0	1	0	0	0	0*	0*	1	0*	1	0*	0*	0	0	5	29
H	Constantinou et al. (2017)	1	1	1	2	1	1	1	0*	1	1	1	1	1	1	0	1	15	88
	Diamond et al. (2020)	1	1	1	2	1	1	0	0*	0*	1	1	1	0*	0*	0	0	10	59
I	Elitzen et al. (2012)	1	1	1	2	1	1	1	1	1	1	1	1	0	0*	1	0	14	82
J	Foucher (2017)	1	1	1	2	1	1	1	0*	0*	1	1	1	1	0*	1	0	13	76
K	Foucher et al. (2007)	1	1	0	2	0	1	1	0*	0*	1	1	1	0*	0*	1	0	10	59

Table 4.4: Results of the risk of bias assessment using the checklist of Downs & Black (1998).

Study	Report	Q1	Q2	Q3	Q5	Q6	Q7	Q10	Q11	Q12	Q16	Q18	Q20	Q21	Q22	Q25	Q27	Total	Total [%]
L	Foucher & Wimmer (2012)	1	1	0	1	0	1	1	0*	0*	1	1	1	0*	0*	0	0	8	47
	Foucher et al. (2011)	1	1	0	2	0	1	1	0*	0*	1	1	1	0*	0*	1	1	11	65
M	Hall et al. (2017)	1	1	1	2	1	1	1	0*	0*	1	1	1	1	0*	0	0	12	71
N	Hara et al. (2016) CON data: Hara et al. (2014)	1	1	1	1	1	1	0	0*	0*	1	1	1	0*	0*	0	0	9	53
O	Hurwitz et al. (1997)	1	0	0	2	1	1	1	0*	0*	0	1	1	0*	0*	1	0	9	53
P	Ismailidis et al. (2021)	1	1	1	2	1	1	1	0*	0*	1	1	1	0*	0*	0	1	12	71
	Ismailidis et al. (2020)	1	1	1	2	1	1	1	0*	0*	1	1	1	0	0*	0	1	12	71
	Nilesch et al. (2021)	1	1	1	2	1	1	1	0*	0*	1	1	1	0*	0*	0	1	12	71
Q	Katoka et al. (2021)	1	1	1	2	1	0	1	0*	0*	1	1	1	0*	0*	1	0	11	65
R	Kubota et al. (2007)	1	1	1	2	1	1	0	0*	0*	1	0	1	0*	0*	1	0	10	59
S	Kumar et al. (2015)	1	1	1	2	1	1	1	0*	0*	1	1	1	1	0*	1	0	13	76
T	Leigh et al. (2016)	1	1	1	2	1	1	1	0*	0*	1	1	1	0*	0*	1	1	13	76

Table 4.4: Results of the risk of bias assessment using the checklist of Downs & Black (1998).

Study	Report	Q1	Q2	Q3	Q5	Q6	Q7	Q10	Q11	Q12	Q16	Q18	Q20	Q21	Q22	Q25	Q27	Total	Total [%]
U	Meyer et al. (2018)	1	1	1	2	1	1	1	0*	0*	1	1	1	0	0*	1	0	12	71
	Wesseling et al. (2018)	1	1	0	2	1	1	1	0*	0*	1	1	1	0*	0*	1	0	11	65
V	Ornetti et al. (2011)	1	1	1	1	1	1	1	0*	0*	1	1	1	0*	0*	0	0	10	59
W	Popovic et al. (2021)	1	1	1	2	1	1	1	0*	0	1	1	1	1	0*	1	0	13	76
X	Porta et al. (2021)	1	1	1	2	1	1	1	0	0*	1	?	1	0	0*	1	0	12	71
Y	Rehnga et al. (2012)	1	1	1	2	1	1	1	0*	0*	1	1	1	0	0*	1	0	12	71
Z	Schmidt et al. (2017)	1	1	1	2	1	1	1	0*	0*	1	1	1	0	0*	0	0	11	65
	Stief et al. (2018)	1	1	1	2	1	1	1	0*	0*	1	1	1	0	0*	0	0	11	65
AA	Schmitt et al. (2015)	1	1	1	2	1	1	1	0*	0*	1	1	1	0	0*	0	0	11	65
AB	Steingrebe et al. (2022)	1	1	1	2	1	1	1	0	0*	1	1	1	0*	0*	1	0	12	71
AC	Tanaka (1998)	0	1	0	1	0	0	0	0*	0*	0	0*	0	0*	0*	0	0	2	12
AD	Tateuchi et al. (2014)	1	1	0	2	1	1	0	0*	0*	1	1	1	0*	0*	0	0	9	53
AE	Thurston (1985)	1	1	0	1	1	1	0	0*	0*	1	0	1	0*	0*	0	0	7	41

Table 4.4: Results of the risk of bias assessment using the checklist of Downs & Black (1998).

Study	Report	Q1	Q2	Q3	Q5	Q6	Q7	Q10	Q11	Q12	Q16	Q18	Q20	Q21	Q22	Q25	Q27	Total	Total [%]
Z & AF	van Drongelen, Braun, et al. (2021)	1	1	0	2	1	1	0	0*	0*	1	1	1	0	0*	0	0	9	53
	van Drongelen, Stetter, et al. (2021)	1	1	0	2	1	1	1	0*	0*	1	1	1	0	0*	0	0	10	59
AF	van Drongelen, Kaldowski, et al. (2020)	1	1	0	2	1	1	1	0*	0*	1	1	1	0*	0*	0	0	10	59
	van Drongelen et al. (2019)	1	1	0	2	1	1	1	0*	0*	1	1	1	1	0*	0	0	11	65
AG	Watanabe et al. (1999)	1	1	0	1	1	1	0	0*	0*	1	1	1	0*	0*	0	0	8	47
	Watanabe et al. (1998)	1	1	0	1	1	1	0	0*	0*	1	1	1	0*	0*	0	0	8	47
AH	Watelein et al. (2001)	1	1	1	2	1	1	0	0*	0*	1	1	1	0*	0*	0	0	10	59
AI	Züchner et al. (2018)	1	1	1	2	1	1	1	0*	0*	1	1	1	0	0*	1	0	12	71

Note. * = unable to determine.

4.4.2 Gait

The results from 33 studies analysing gait movement are reviewed below. The results are presented separately by joint, movement plane and laterality. Within each subsection, the results are presented in the following order: results of the meta-analysis, a qualitative summary of data and studies not included in the meta-analysis, results of continuous angle-time curve analyses and, if available, results of subgroup analyses.

4.4.2.1 Hip joint

4.4.2.1.1 Sagittal plane kinematics

Ipsilateral

Individuals with HOA have reduced peak extension of the affected hip joint during the gait cycle (GC) (10 studies, SMD = -1.28; 95%CI = -1.61, -0.95; $I^2 = 28$; Figure 4.2A). Similar results were found for the peak hip extension during ST (7 studies, SMD = -1.22; 95%CI = -1.72, -0.71; $I^2 = 71$; Figure 4.2B). Hip sagittal angle at toe-off (TO) was significantly reduced (5 studies, SMD = -0.86; 95%CI = -1.41, -0.32; $I^2 = 45$; Figure 4.3C). Peak flexion was not different between HOA and CON participants (GC: 8 studies, SMD = -0.53; 95%CI = -1.25, 0.19; $I^2 = 80$; ST: 5 studies, SMD = -0.36; 95%CI = -1.34, 0.62; $I^2 = 86$; swing phase: 2 studies, SMD = -1.26; 95%CI = -6.81, 4.28; $I^2 = 54$).

Hip flexion at initial contact (IC) did not differ between groups (3 studies, SMD = -0.67; 95%CI = -2.09, 0.76; $I^2 = 76$).

There was a significant reduction of the sagittal hip RoM across GC (9 studies, SMD = -2.80; 95%CI = -3.84, -1.75; $I^2 = 91$, Figure 4.3D), but not across ST (5 studies, SMD = -1.42; 95%CI = -3.15, 0.31; $I^2 = 95$).

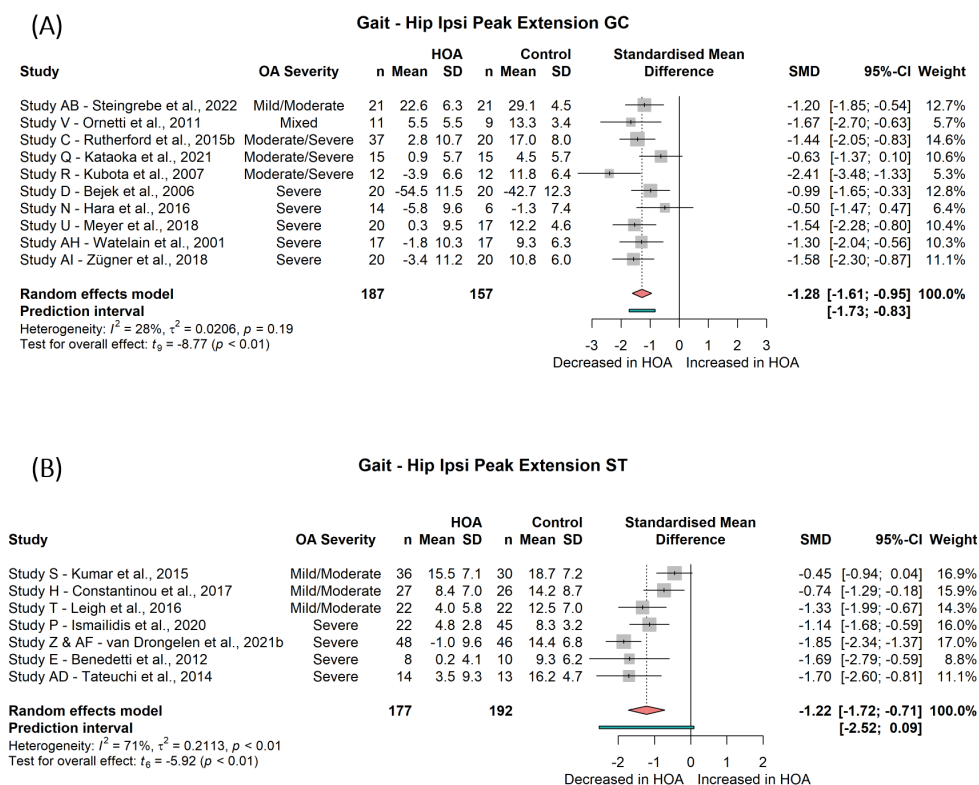


Figure 4.2: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for: (A) ipsilateral peak hip extension during gait cycle (GC), (B) ipsilateral peak hip extension during stance phase (ST); during gait.

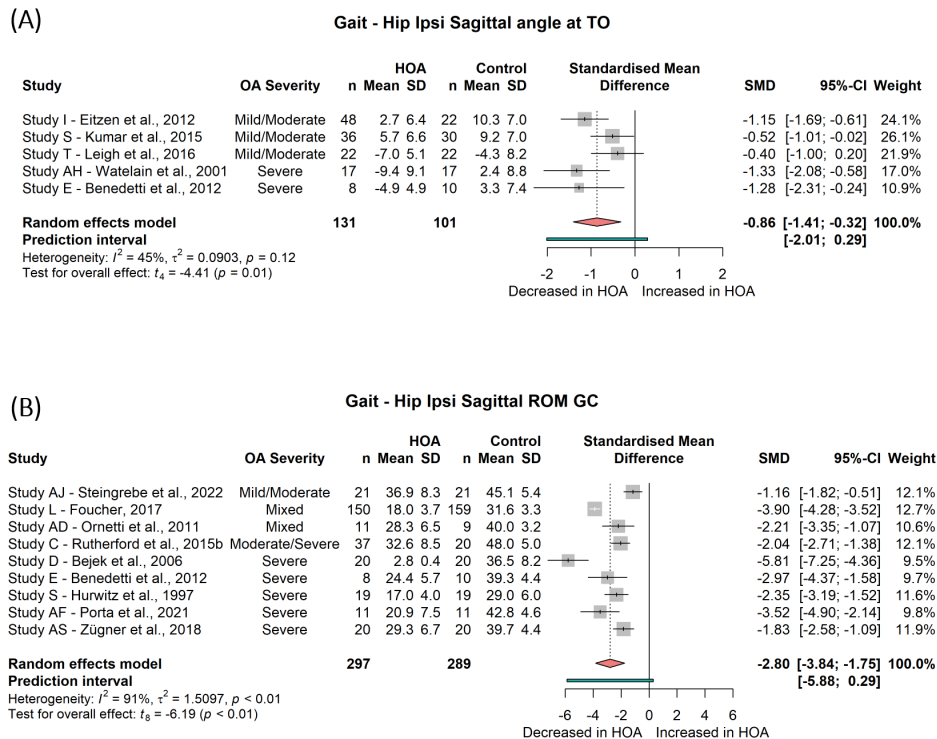


Figure 4.3: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for: (A) ipsilateral hip sagittal angle at toe-off (TO), (B) ipsilateral hip sagittal range of motion (RoM) across the gait cycle (GC); during gait.

Studies not included in the meta-analysis also showed reduced peak hip extension (Schmitt et al., 2015; Tanaka, 1998). Results on peak hip flexion varied, with Schmitt et al. (2015) reporting increased peak hip flexion in individuals with unilateral HOA and Tanaka (1998) in individuals with bilateral HOA. In contrast, Tanaka (1998) showed reduced peak hip flexion in individuals with unilateral HOA. Additionally, a reduction of hip flexion at IC was found (Schmitt et al., 2015). Reduced sagittal hip RoM was found in three out of four studies (significant reduction in RoM during ST (Foucher et al., 2007; Foucher & Wimmer, 2012) and swing phase (Ismailidis et al., 2020); no significant difference in RoM during GC (Brand & Crowninshield, 1980).

All six studies (Ardestani & Wimmer, 2016; Study P - Ismailidis et al., 2021; Study U - Meyer et al., 2018; Study P - Nüesch et al., 2021; Porta et al., 2021; Study Z & AF - van Drongelen, Braun, et al., 2021; Study Z & AF - van Drongelen, Stetter, et al., 2021; Study U - Wesseling et al., 2018) analysing continuous sagittal hip angle-time curves show differences between HOA and CON participants.

Subgroup analysis did not find significant differences in peak hip flexion during ST or GC for individuals with mild/moderate HOA (ST: 2 studies, SMD = 0.31; 95%CI = -4.33, 4.95; $I^2 = 86$; GC: 2 studies, SMD = -0.36; 95%CI = -1.53, 0.82 $I^2 = 0$) or individuals with severe HOA (ST: 3 studies, SMD = -0.80; 95%CI = -2.28, 0.67; $I^2 = 80$; GC: 4 studies, SMD = -0.61; 95%CI = -2.61, 1.39; $I^2 = 85$).

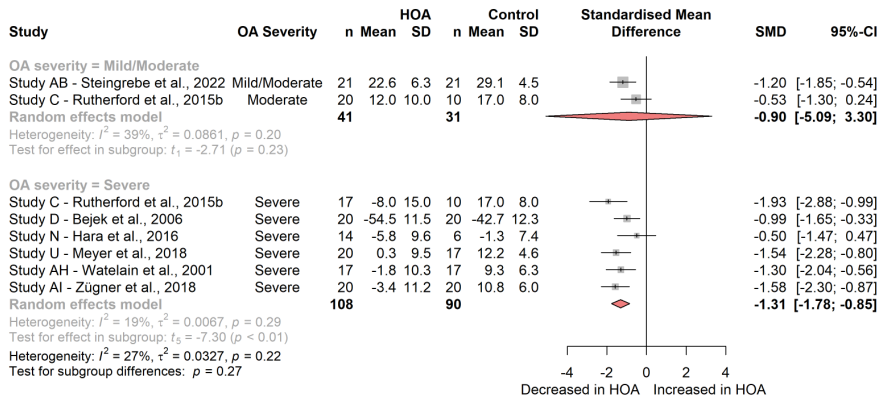
Hip peak extension during ST or GC was significantly different for individuals with severe HOA (ST: 4 studies, SMD = -1.57; 95%CI = -2.14, -1.00; $I^2 = 22$; GC: 6 studies, SMD = -1.31; 95%CI = -1.78, -0.85; $I^2 = 19$, Figure 4.4) but not for individuals with mild/moderate HOA (ST: 3 studies, SMD = -0.80; 95%CI = -1.89, 0.29; $I^2 = 55$; GC: 2 studies, SMD = -0.90; 95%CI = -5.09, 3.30; $I^2 = 39$).

Hip sagittal angle at TO was significantly decreased in severe HOA (2 studies, SMD = -1.31; 95%CI = -1.63, -0.99; $I^2 = 0$, Figure 4.5B), but not in mild/moderate HOA (3 studies, SMD = -0.69; 95%CI = -1.68, 0.30; $I^2 = 52$).

Hip sagittal RoM across GC was significantly decreased for mild/moderate (2 studies, SMD = -1.19; 95%CI = -1.71, -0.68; $I^2 = 0$) and severe HOA (6 studies, SMD = -3.23; 95%CI = -4.65, -1.80; $I^2 = 81$) (Figure 4.5A). Hip sagittal RoM across ST was not different for any of the HOA severity subgroups (mild/moderate: 3 studies, SMD = -0.70; 95%CI = -2.85, 1.45; $I^2 = 91$; severe: 2 studies, SMD = -2.51; 95%CI = -16.52, 11.51; $I^2 = 96$).

(A)

SUB - Gait - Hip Ipsi Peak Extension GC



(B)

SUB - Gait - Hip Ipsi Peak Extension ST

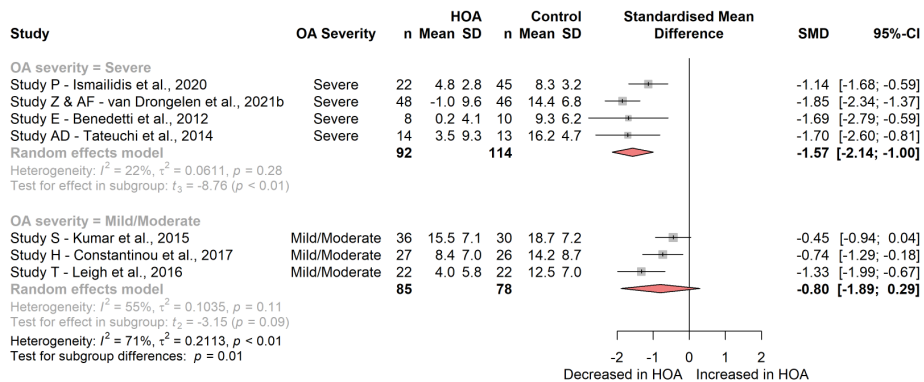
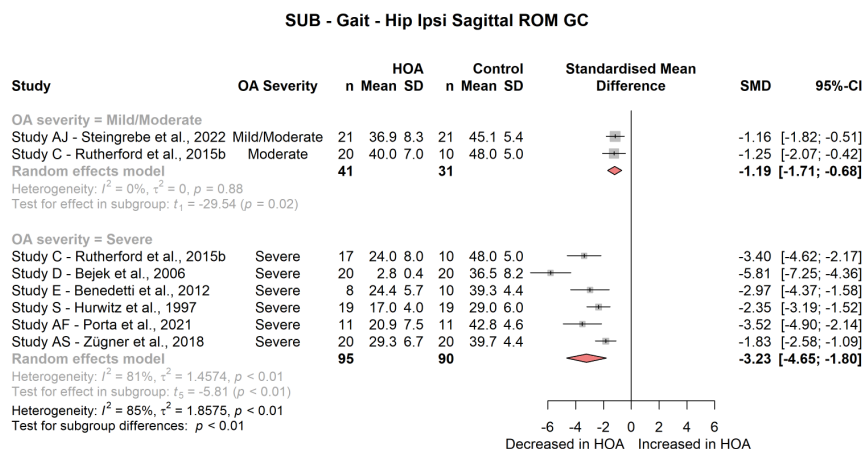


Figure 4.4: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for subgroup analyses on: (A) ipsilateral peak hip extension during gait cycle (GC), (B) ipsilateral peak hip extension during stance phase (ST); during gait.

(A)



(B)

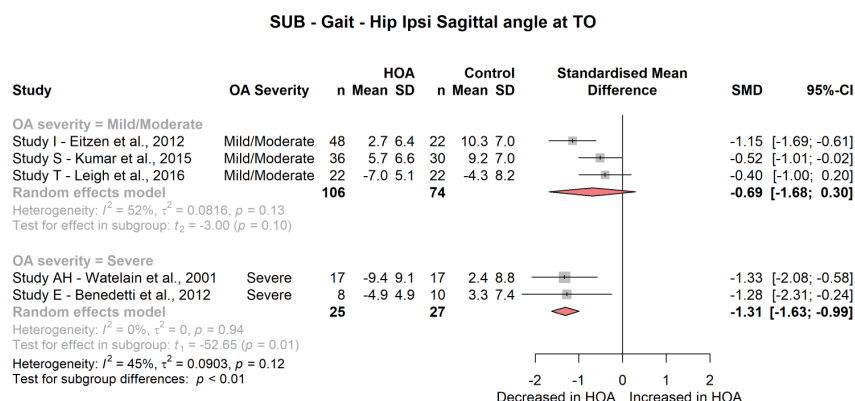


Figure 4.5: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for subgroup analyses on: (A) ipsilateral hip sagittal range of motion (RoM) across gait cycle (GC), (B) ipsilateral hip sagittal angle at toe-off (TO); during gait.

Contralateral

Individuals with unilateral HOA also displayed a reduced peak extension of the contralateral hip joint during ST of gait (3 studies, SMD = -0.59; 95%CI = -0.97, -0.22; $I^2 = 0$, Figure 4.6) but not during GC (2 studies, SMD = -0.15; 95%CI = -7.76, 7.45; $I^2 = 78$). No differences between HOA and CON participants were found for contralateral peak hip flexion during ST (3 studies, SMD = 0.23; 95%CI = -0.41, 0.87; $I^2 = 0$) or GC (2 studies, SMD = -0.7; 95%CI = -2.38, 0.98; $I^2 = 0$), nor for contralateral sagittal hip RoM across ST (3 studies, SMD = -0.3; 95%CI = -0.74, 0.15; $I^2 = 0$) or GC (4 studies, SMD = -0.79; 95%CI = -2.65, 1.06; $I^2 = 88$) during the meta-analysis.

A study not included in the meta-analysis also reported reduced contralateral peak hip extension and increased peak hip flexion angles (Tanaka, 1998).

Of the four studies (Nüesch et al., 2021; Porta et al., 2021; van Drongelen, Braun, et al., 2021; Wesseling et al., 2018) analysing contralateral sagittal hip angle-time curves, three found differences in contralateral hip sagittal angles.

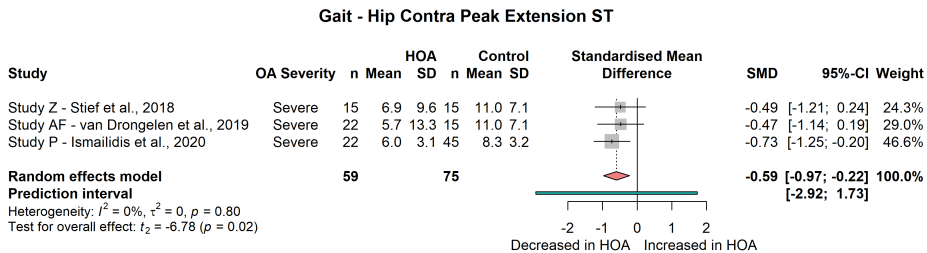


Figure 4.6: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for contralateral peak hip extension during stance phase (ST); during gait.

4.4.2.1.2 Frontal plane kinematics

Ipsilateral

The meta-analysis showed no significant differences for peak hip abduction during GC (4 studies, SMD = -0.7; 95%CI = -1.83, 0.44; $I^2 = 75$) or at TO (2 studies, SMD = -0.16; 95%CI = -5.24, 4.92; $I^2 = 66$). Peak hip adduction did not differ significantly between groups either during GC or ST (GC: 5 studies, SMD = -0.35; 95%CI = -0.87, 0.18; $I^2 = 42$; ST: 4 studies, SMD = -0.43; 95%CI = -1.24, 0.38; $I^2 = 46$), but a significant reduction of the frontal plane hip

RoM across GC (4 studies, SMD = -0.86; 95%CI = -1.93, -0.33; $I^2 = 0$, Figure 4.7A) was found. Frontal plane hip RoM across ST was not different between groups (2 studies, SMD = -1.38; 95%CI = -11.65, 8.88; $I^2 = 88$).

Studies not included in the meta-analysis showed an increased hip abduction at midstance (Leigh et al., 2016), as well as decreased peak hip adduction during early and late ST (Diamond et al., 2020). Peak abduction during ST (Tateuchi et al., 2014) and swing (Benedetti et al., 2012), as well as hip frontal angle at peak hip extension (Leigh et al., 2016), did not differ between subject groups.

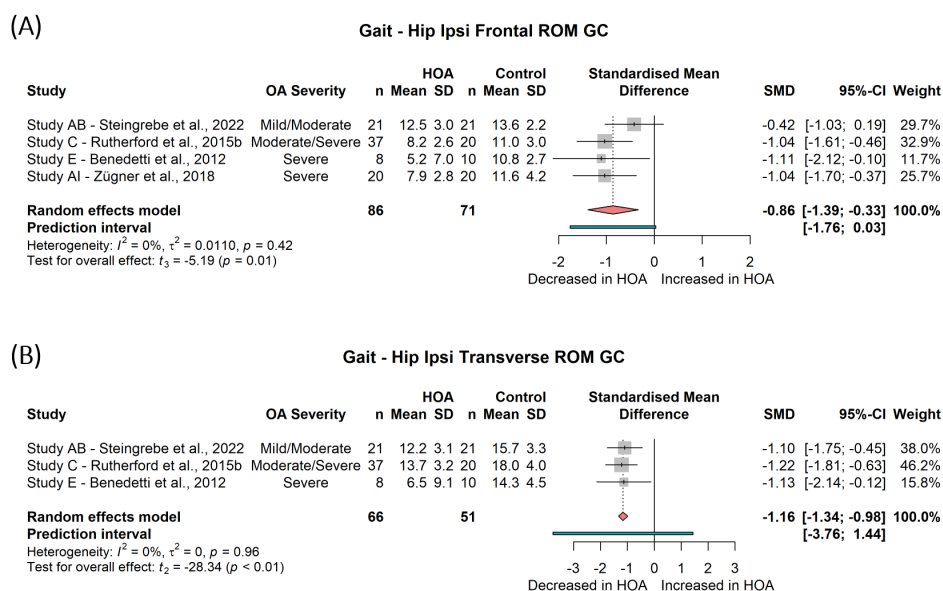


Figure 4.7: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) ipsilateral hip frontal range of motion (RoM) across gait cycle (GC), (B) ipsilateral hip transverse RoM across GC; during gait.

Both studies using statistical parametric mapping (SPM) analysis found differences in hip frontal plane angle (Study U - Meyer et al., 2018; van Drongelen, Braun, et al., 2021; Study U - Wesseling et al., 2018). van Drongelen, Braun, et al. (2021) found differences for individuals with unilateral HOA but not for those with bilateral HOA.

Results from the subgroup analysis did not find group differences in peak hip abduction or adduction during GC for mild/moderate HOA (abduction: 2 studies, SMD = -0.01; 95%CI = -2.05, 2.03; $I^2 = 0$; adduction: 2 studies, SMD = -0.44; 95%CI = -2.80, 1.91; $I^2 = 0$) or for severe HOA (abduction: 2 studies, SMD = -0.94; 95%CI = -7.33, 5.46; $I^2 = 70$; adduction: 2 studies, SMD = -0.02; 95%CI = -7.02, 6.98; $I^2 = 78$).

Hip frontal RoM across GC was not different in mild/moderate (2 studies, SMD = -0.39; 95%CI = -0.93, 0.16; $I^2 = 0$) or in severe HOA (2 studies, SMD = -1.35; 95%CI = -2.74, 0.04; $I^2 = 37$).

Contralateral

van Drongelen et al. (2019; 2020) found no significant difference between CON and HOA groups for the peak adduction angle during ST.

Results from two SPM analyses (van Drongelen, Braun, et al., 2021; Wesseling et al., 2018) show differences in contralateral hip frontal angle-time curves.

4.4.2.1.3 Transverse plane kinematics

Ipsilateral

The meta-analysis yielded no difference for the peak external rotation angle (2 studies, SMD = 0.17; 95%CI = -0.59, 0.94; $I^2 = 0$) or the peak internal rotation angle during GC (3 studies, SMD = -0.5; 95%CI = -1.75, 0.76; $I^2 = 59$). Also, the transverse angle at TO was not different (2 studies, SMD = -0.11; 95%CI = -14.61, 14.39; $I^2 = 95$). However, a significantly decreased transverse RoM across GC (3 studies, SMD = -1.16; 95%CI = -1.34, -0.98; $I^2 = 0$, Figure 4.7B) was found.

Data not included in the meta-analysis showed a significant reduction of the peak internal rotation angle during ST and a significant increase in peak external rotation during the swing phase (Benedetti et al., 2012). Similarly, Leigh et al. (2016) found that the hip joint of individuals with HOA was significantly more externally rotated at terminal hip extension but not at midstance.

One study (Wesseling et al., 2018) found differences in the hip transverse angle-time curve using SPM analysis.

The subgroup analysis did not show significant differences in hip transverse RoM across GC for mild/moderate (2 studies, SMD = -0.84; 95%CI = -4.60, 2.93; $I^2 = 26$) or severe HOA (2 studies, SMD = -1.47; 95%CI = -5.39, 2.45; $I^2 = 0$).

Contralateral

Only one study (Wesseling et al., 2018) reported data on transverse plane contralateral hip angles and found no significant differences during an SPM analysis.

4.4.2.2 Knee joint

4.4.2.2.1 Sagittal plane kinematics

Ipsilateral

The meta-analysis did not reveal a significant difference for peak knee extension during GC (3 studies, SMD = 0.72; 95%CI = -2.21, 3.65; $I^2 = 87$) or ST (2 studies, SMD = 0.81; 95%CI = -13.61, 15.23; $I^2 = 96$). Peak flexion was significantly reduced during GC (3 studies, SMD = -0.87; 95%CI = -1.19, -0.56; $I^2 = 0$, Figure 4.8A), but not during ST (3 studies, SMD = -0.68; 95%CI = -1.55, 0.2; $I^2 = 44$). Sagittal knee angle at IC (2 studies, SMD = -0.09; 95%CI = -0.54, 0.37; $I^2 = 0$), midstance (2 studies, SMD = 0.22; 95%CI = -4.26, 4.69; $I^2 = 68$), TO (2 studies, SMD = 0.22; 95%CI = -1.60, 2.03; $I^2 = 0$) and peak hip extension (2 studies, SMD = 1.30; 95%CI = -0.52, 3.11; $I^2 = 0$) was not different between groups. Sagittal knee RoM was significantly reduced across GC (4 studies, SMD = -1.64; 95%CI = -2.24, -0.86; $I^2 = 5$, Figure 4.8B) but not across ST (3 studies, SMD = -1.06; 95%CI = -5.25, 3.13; $I^2 = 97$).

A study not included in the meta-analysis by Ismailidis et al. (2020), described a significantly decreased knee RoM across the swing phase, while peak flexion during swing did not differ between groups.

Analyses of knee sagittal angle-time curves showed differences in all five studies (Ardestani & Wimmer, 2016; Study P - Ismailidis et al., 2021; Study P - Nüesch et al., 2021; Porta et al., 2021; Study Z & AF - van Drongelen, Braun, et al., 2021; Study Z & AF - van Drongelen, Stetter, et al., 2021; Wesseling et al., 2018).

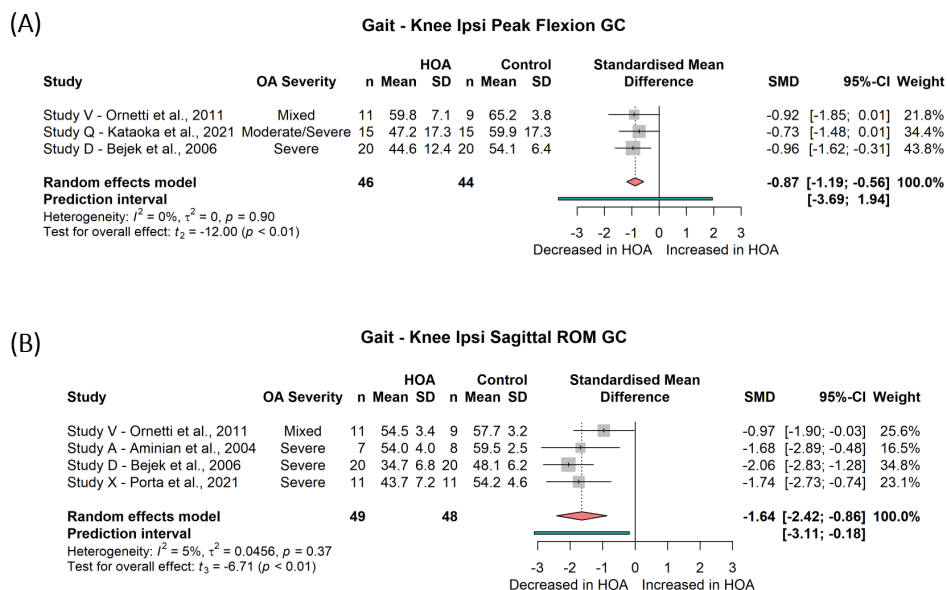


Figure 4.8: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for: (A) ipsilateral knee peak flexion during gait cycle (GC), (B) ipsilateral knee sagittal range of motion (RoM) across GC; during gait.

Contralateral

For the contralateral knee joint, the meta-analysis did not show a significant difference for peak knee extension during GC (2 studies, SMD = -0.03; 95%CI = -1.77, 1.71; $I^2 = 0$) but a significant difference during ST (2 studies, SMD = 0.52; 95%CI = 0.18, 0.87; $I^2 = 0$, Figure 4.9A). Peak flexion was not significantly reduced during GC (2 studies, SMD = -0.87; 95%CI = -3.21, 1.47; $I^2 = 0$) or ST (3 studies, SMD = -0.05; 95%CI = -0.66, 0.55; $I^2 = 0$). Sagittal knee RoM was significantly reduced across both, GC (5 studies, SMD = -0.73; 95%CI = -1.08, -0.39; $I^2 = 0$, Figure 4.9B) and ST (3 studies, SMD = -0.65; 95%CI = -0.71, -0.59; $I^2 = 0$, Figure 4.9C).

Out of four studies analysing sagittal contralateral knee angles time curves, two found differences between subject groups (Porta et al., 2021; Wesseling et al., 2018) and two did not (Nüesch et al., 2021; van Drongelen, Braun, et al., 2021).

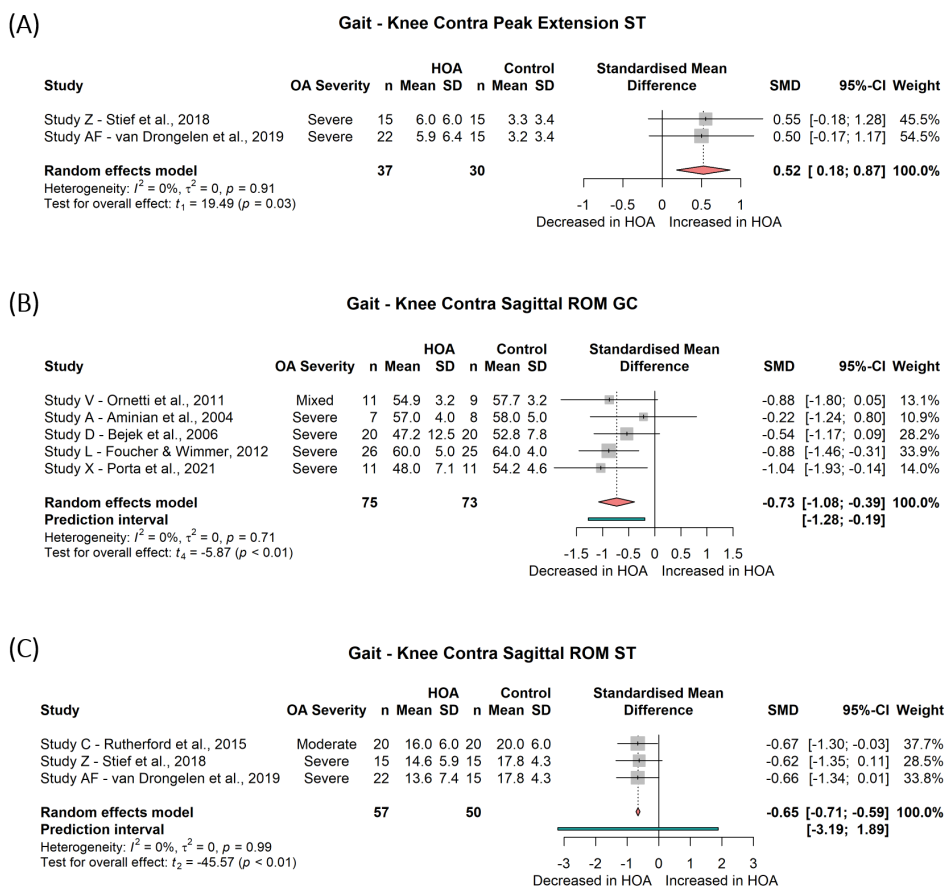


Figure 4.9: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for: (A) contralateral knee peak extension during stance phase (ST), (B) contralateral knee sagittal range of motion (RoM) across gait cycle (GC), (C) contralateral knee sagittal RoM across ST; during gait.

4.4.2.2.2 Frontal plane kinematics

Ipsilateral

The meta-analysis did not show a difference between groups for the RoM across ST (2 studies, $SMD = -0.09$; $95\%CI = -5.55, 5.37$; $I^2 = 68$).

One study not included in the meta-analysis reported no significant group differences for the frontal knee angle at midstance, TO or peak hip extension (Leigh et al., 2016)

Contralateral

The meta-analysis did not show a difference between groups for the RoM across ST (2 studies, SMD = -0.08; 95%CI = -5.67, 5.51; $I^2 = 69$).

4.4.2.2.3 Transverse plane kinematics

Ipsilateral

Rutherford et al. (2015a) did not find a significant difference in transverse knee RoM across ST. However, Leigh et al. (2016) found significantly increased external knee rotation angles at midstance and at peak hip extension but not at TO.

Contralateral

No studies were retrieved that presented data on contralateral transverse plane knee kinematics.

4.4.2.3 Ankle joint

4.4.2.3.1 Sagittal plane kinematics

Ipsilateral

The meta-analysis did not show differences for the peak dorsiflexion angle during GC (2 studies, SMD = -0.01; 95%CI = -9.38, 9.37; $I^2 = 82$) or ST (2 studies, SMD = -0.62; 95%CI = -3.77, 5.00; $I^2 = 54$). Peak plantar flexion during GC (3 studies, SMD = 0.46; 95%CI = -0.98, 1.90; $I^2 = 51$) and during ST (2 studies, SMD = -0.09; 95%CI = -3.67, 3.49; $I^2 = 35$) was not different between groups. Ankle sagittal RoM across GC (2 studies, SMD = -0.53; 95%CI = -6.47, 5.41; $I^2 = 53$) and across ST (2 studies, SMD = 0.34; 95%CI = -1.61, 2.29; $I^2 = 0$) did not differ between groups. The ankle angles at IC (2 studies, SMD = 0.07; 95%CI = -2.45, 2.58; $I^2 = 25$), midstance (2 studies, SMD = 0.18; 95%CI = -1.24, 1.60; $I^2 = 0$), TO (2 studies, SMD = 0.02; 95%CI = -1.42, 1.47; $I^2 = 0$) and peak hip extension (2 studies, SMD = 0.25; 95%CI = -1.72, 2.21; $I^2 = 0$) were not different between groups.

In five studies, ankle sagittal angle-time curves were analysed. Although two studies found differences between groups (Porta et al., 2021; Wesseling et al., 2018), two did not (Ardestani & Wimmer, 2016; van Drongelen, Braun, et al., 2021) and one study yielded contradicting results in two reports (Study P - Ismailidis et al., 2021; Study P - Nüesch et al., 2021).

Contralateral

Ankle angle RoM across GC was not different between groups (2 studies, SMD = -0.47; 95%CI = -3.88, 2.94; $I^2 = 0$).

Data not included in the meta-analysis showed differences in peak dorsiflexion angle during GC ($p = 0.05$) but not in peak plantar flexion angle ($p = 0.087$) (Ornetti et al., 2011).

In four studies, contralateral sagittal ankle angle-time curves were analysed. Three of these studies found differences between groups (Nüesch et al., 2021; Porta et al., 2021; Wesseling et al., 2018) but one study did not (van Drongelen, Braun, et al., 2021).

4.4.2.3.2 Frontal plane kinematics

Ipsilateral

Only one study analysed frontal plane ankle angles and found a reduced ankle inversion at TO. Frontal ankle angle at midstance and at peak hip extension did not differ between groups (Leigh et al., 2016).

Contralateral

No studies were retrieved that presented data on contralateral frontal plane ankle kinematics.

4.4.2.3.3 Transverse plane kinematics

Ipsilateral

Only one study analysed transverse plane ankle kinematics and did not find differences in ankle angles at midstance, TO or peak hip extension (Leigh et al., 2016).

Contralateral

No studies were retrieved that presented data on contralateral transverse plane ankle kinematics.

4.4.2.4 Pelvis

4.4.2.4.1 Sagittal plane kinematics

The meta-analysis found no differences in peak anterior tilt during either GC (4 studies, SMD = 0.70; 95%CI = -0.11, 1.51; $I^2 = 27$) or ST (2 studies, SMD = 0.68; 95%CI = -5.07, 6.43; $I^2 = 85$). Likewise, pelvis peak posterior tilt did not differ between groups during GC (3 studies, SMD = 0.72; 95%CI = -1.78, 3.23; $I^2 = 79$). Pelvis angle at TO also did not differ between CON and HOA participants (2 studies, SMD = -0.38; 95%CI = -12.24, 11.49; $I^2 = 93$). Pelvis RoM across GC was not different between groups (3 studies, SMD = 1.85; 95%CI = -1.99, 5.69; $I^2 = 91$).

Studies not included in the meta-analysis found a significantly increased anterior pelvis tilt at peak hip extension (Leigh et al., 2016), but not at midstance (Leigh et al., 2016; Watanabe et al., 1999), IC or TO (Watanabe et al., 1999) in individuals with HOA.

One study analysed pelvis sagittal angle using SPM analysis and found differences between individuals with bilateral HOA and CON, but not between individuals with unilateral HOA and CON (van Drongelen, Braun, et al., 2021).

4.4.2.4.2 Frontal plane kinematics

The meta-analysis found no differences in peak inferior obliquity during GC (2 studies, SMD = 2.25; 95%CI = -12.44, 16.93; $I^2 = 92$). Peak superior obliquity was not different in GC (2 studies, SMD = -0.28; 95%CI = -7.83, 7.27; $I^2 = 80$) or in ST (3 studies, SMD = -0.43; 95%CI = -2.53, 1.61; $I^2 = 78$). Likewise, pelvic frontal angle at TO did not differ between groups (2 studies, SMD = -0.53; 95%CI = -8.60, 7.54; $I^2 = 86$).

Pelvis frontal plane RoM was not different between groups either across GC (5 studies, SMD = -0.19; 95%CI = -3.33, 2.96; $I^2 = 96$) or ST (2 studies, SMD = -0.31; 95%CI = -6.80, 6.18; $I^2 = 81$).

Studies not included in the meta-analysis found no differences in peak inferior obliquity during the swing phase (Benedetti et al., 2012); or for the pelvis frontal angle at peak hip extension (Leigh et al., 2016), IC, midstance or TO (study AG: Watanabe et al., 1998, 1999).

Differences between groups were found for pelvis frontal angle at midstance (Leigh et al., 2016) and peak inferior obliquity during single-limb stance (van Drongelen, Kaldowski, et al., 2020).

One study analysed pelvis frontal angle using SPM analysis and found differences between bilateral HOA and CON participants, but not between unilateral HOA and CON participants (van Drongelen, Braun, et al., 2021).

4.4.2.4.3 Transverse plane kinematics

The meta-analysis did not show differences in the pelvis transverse RoM across GC (3 studies, SMD = -0.06; 95%CI = -0.44, 0.32; $I^2 = 0$) or the pelvis transverse angle at TO (2 studies, SMD = 0.35; 95%CI = -3.13, 3.82; $I^2 = 29$).

Studies not included in the meta-analysis found no differences between groups for the transverse pelvic RoM across ST (Watelain et al., 2001), pelvis angle at peak hip extension (Leigh et al., 2016) or peak posterior rotation (Benedetti et al., 2012). However, one study reported significant group differences in the pelvis transverse angle at midstance (Leigh et al., 2016).

4.4.3 Stair walking

Two studies analysing stair walking are reviewed below, separated into stair ascent and stair descent.

4.4.3.1 Stair ascent

Peak ipsilateral hip flexion, adduction and internal rotation during ST did not differ between groups (flexion: 2 studies, SMD = -0.49; 95%CI = -9.71, 8.73; $I^2 = 90$; adduction: 2 studies, SMD = -0.17; 95%CI = -8.42, 8.08; $I^2 = 88$; internal rotation: 2 studies, SMD = 0.41; 95%CI = -3.03, 3.85; $I^2 = 38$).

Results not included in the meta-analysis showed no differences in peak hip extension during GC (Hall et al., 2017) or during ST (Popovic et al., 2021). Peak flexion was significantly reduced in the swing phase (Hall et al., 2017). Hip sagittal RoM across GC was significantly reduced (Hall et al., 2017).

Hip peak abduction was significantly reduced during swing phase (Hall et al., 2017) and ST (Popovic et al., 2021). Hip frontal RoM was significantly reduced (Hall et al., 2017).

Peak external rotation was significantly reduced during ST (Popovic et al., 2021) but not during swing phase (Hall et al., 2017). Transverse hip RoM was significantly reduced (Hall et al., 2017).

Peak ipsilateral knee angles in the sagittal and frontal planes during ST did not differ between groups (Popovic et al., 2021). Likewise, peak knee internal rotation was not different. However, peak knee external rotation was significantly increased in individuals with HOA.

Peak ipsilateral sagittal and frontal ankle angles did not differ between groups. However, peak ankle internal rotation was significantly reduced, while peak external rotation was significantly increased, in individuals with HOA (Popovic et al., 2021).

Peak contralateral pelvis inferior and superior obliquity did not differ between groups (Hall et al., 2017).

4.4.3.2 Stair descent

Peak ipsilateral hip flexion during ST did not differ between groups (2 studies, SMD = -0.06; 95%CI = -1.36, 1.25; $I^2 = 0$). Likewise, peak hip adduction and abduction did not differ between CON and HOA participants (adduction: 2 studies, SMD = 0.05; 95%CI = -5.10, 5.21; $I^2 = 71$; abduction: 2 studies, SMD = -0.03; 95%CI = -5.47, 5.42; $I^2 = 74$). Peak internal and external hip rotation were not different between groups (internal rotation: 2 studies, SMD = 0.15; 95%CI = -4.21, 4.52; $I^2 = 60$; external rotation: 2 studies, SMD = -0.37; 95%CI = -1.42, 0.69; $I^2 = 0$).

Results not included in the meta-analysis showed no differences in peak hip extension during either ST (Popovic et al., 2021) or GC (Hall et al., 2017). Likewise, peak flexion during swing was not different between groups (Hall et al., 2017). However, sagittal hip RoM was significantly reduced in individuals with HOA (Hall et al., 2017).

There was no difference between groups in peak hip abduction during swing, hip frontal plane RoM across GC or transverse plane hip RoM (Hall et al., 2017).

While peak knee flexion did not differ between groups, peak knee extension was significantly increased in individuals with HOA (Popovic et al., 2021).

Frontal plane peak knee angles did not differ between groups (Popovic et al., 2021).

Transverse plane knee angles showed significantly reduced peak internal rotation and significantly increased peak external rotation (Popovic et al., 2021).

At the ankle joint, individuals with HOA showed significantly increased peak plantar flexion. Peak dorsiflexion did not differ between groups, and neither did peak ankle eversion and inversion (Popovic et al., 2021).

Ankle peak internal rotation was significantly reduced while peak external rotation was significantly increased in individuals with HOA (Popovic et al., 2021).

Peak contralateral pelvis inferior and superior obliquity did not differ between groups (Hall et al., 2017).

4.4.4 Turning while walking

In a study by Tateuchi et al. (2014), 45° turns conducted either in a step turn or in a spin turn manner were analysed in individuals with severe HOA. Peak angles during ST were described.

During the step turn, significantly decreased peak hip flexion and extension were found. Peak hip abduction was also significantly reduced. No differences were found for peak hip adduction.

Sagittal plane peak angles of the knee and ankle joint did not differ between groups.

During the spin turn, significantly decreased peak hip flexion and extension were found. Peak hip abduction did not differ between groups. However, significantly reduced peak hip adduction was found in individuals with HOA.

Peak knee extension did not differ between groups, but peak knee flexion was significantly reduced in individuals with HOA.

Ankle sagittal peak angles did not differ between groups.

4.5 Discussion

The aim of this review and meta-analysis was to summarise existing literature on lower-limb joint kinematics during locomotion in individuals with HOA compared to CON. Where possible, a meta-analysis was performed with the focus on HOA severity and uni- or bilateral involvement.

Overall, 47 reports from 35 individual studies were reviewed. The total score regarding risk of bias and quality of reporting of the included studies varied strongly, with older reports tending to show lower scores.

The first outcome of this systematic literature review is that studies on locomotion tasks other than gait are rare, with only two studies on stair walking (Hall et al., 2017; Popovic et al., 2021) and one study on curve walking (Tateuchi et al., 2014).

Secondly, a large portion of the study participants are classified as having severe HOA as well as unilateral HOA. This observation may originate in the fact that individuals with HOA are often recruited prior to THR, and studies aim at evaluating rehabilitation after surgery. Due to the limited number of studies involving individuals with mild or moderate HOA, subgroup analyses for HOA severity were only possible for certain parameters of ipsilateral hip kinematics. Likewise, it was not possible to perform a subgroup analysis regarding HOA laterality, as data from at least two studies for each subgroup, namely unilateral and bilateral HOA, were not available for any of the revised parameters. This was mainly caused by a lack of bilateral HOA subject groups or the unavailability of laterality information. As no conjoint analysis was possible, insights on the impact of laterality still have to be based on individual study results, such as those of Tanaka (1998) or van Drongelen, Braun, et al. (2021).

Generally, it is worth noting that although 33 individual studies were included on gait, a conjoint analysis is hindered by the multitude of calculated parameters. For example, calculating peak angles or RoM across ST is not comparable to the same parameters calculated across GC. Therefore, of the 68 combined analyses calculated for parameters on gait kinematics, only ten include five or more individual studies. Thereof, seven refer to ipsilateral hip sagittal angles, one refers to ipsilateral hip frontal angles, contralateral knee sagittal angle and pelvis frontal angle, respectively.

In five studies, continuous angle-time curves were analysed by means of an SPM analysis (Ardestani & Wimmer, 2016; Study P - Ismailidis et al., 2021; Study U - Meyer et al., 2018; Study P - Nüesch et al., 2021; van Drongelen, Braun, et al., 2021; Study U - Wesseling et al., 2018) or by point-by-point analysis (Porta et al., 2021). While these approaches can be advantageous, especially during explorative data analysis, the aggregation of results across multiple studies is difficult.

Five studies assessed kinematics using IMU sensors (Aminian et al., 2004; Bolink et al., 2015; Study P - Ismailidis et al., 2020; Kataoka et al., 2021; Study P - Nüesch et al., 2021; Reininga et al., 2012). Most of these studies focused on sagittal hip, knee and ankle angles or frontal pelvis angles and demonstrated significant group differences. Only one study used IMUs on frontal and transverse hip angles (Kataoka et al., 2021), but it did not find significant differences.

4.5.1 Effects of hip osteoarthritis on gait kinematics

For the gait movement, the main results for the hip joint were an overall reduced peak extension in individuals with HOA. However, during the subgroup analysis, only individuals with severe HOA demonstrated significant group differences. Similar results were found for the hip sagittal angle at TO, which is expected since peak hip extension and TO occur very close to each other.

Reduction of the sagittal hip RoM across GC occurred with a very large effect size, and was present in individuals with both mild/moderate as well as severe HOA. In contrast, sagittal hip RoM across ST was not different between groups. Thus, it might be crucial to capture ST and swing phase kinematics to discover deviations in gait caused by HOA.

Interestingly, peak hip extension was also reduced in the contralateral limb in individuals with severe unilateral HOA. As hip extension is closely connected to step length (Wonsetler et al., 2018), this contradicts the consideration of increased step length in the contralateral limb to compensate for decreased ipsilateral step length (Constantinou et al., 2014). However, this may be because in most studies healthy CON participants are not evaluated radiographically, so structural changes in the contralateral limb cannot be excluded.

Frontal as well as transverse plane hip RoMs were reduced across GC. As neither of the peak angles demonstrated significant group differences, the RoM might be more sensitive to group differences as it captures changes occurring at both extrema of the dynamic movement. Little is known about frontal and transverse hip kinematics of the contralateral limb; however, in two studies, analyses of contralateral frontal hip angle-time curves found differences between HOA and CON groups, which require further investigation.

Ipsilateral as well as contralateral sagittal plane knee kinematics are influenced by HOA. The ipsilateral limb shows decreased peak flexion and a reduced sagittal RoM across GC. The contralateral limb demonstrates increased peak extension, but with a negligible effect size. However, sagittal knee RoM across ST and GC was significantly decreased with a small to moderate effect size.

Very few studies analysed the effect of HOA on knee frontal and transverse kinematics. Although no differences were found for frontal plane knee angles, individual study results on knee transverse kinematics varied. Some studies found increased foot-progression angles

during walking for individuals with HOA (Schmidt et al., 2017; Tateuchi et al., 2014), which might be linked to changes in knee rotation.

The meta-analysis did not yield any significant differences in ipsi- or contralateral sagittal ankle kinematics. However, individual study results as well as analysis of angle-time curves partly yielded significant group differences. Thus, ankle kinematics might only differ in specific settings or groups: this should be considered in future studies. For the frontal and transverse planes, no data exist regarding contralateral ankle kinematics. Ipsilateral ankle kinematics in those planes were only analysed in one study, which only found a reduced ankle inversion at TO.

The meta-analysis did not yield any significant differences in pelvis sagittal, frontal or transverse movement. However, it has to be noted that large study heterogeneity was observed for all analysed parameters, especially for the sagittal and frontal planes. For example, in the analysis of the frontal pelvis RoM across GC, the results from Bejek et al. (2006) differed dramatically from those of the other studies and, if excluded, the random-effects model approached statistical significance ($p = 0.06$). One possible explanation for this deviation might come from different measurement techniques, as Bejek and colleagues were the only ones to use an ultrasound-based motion capture system. Peak anterior pelvis tilt approached significance ($p = 0.07$) in the meta-analysis with a moderate effect size. Anterior tilting of the pelvis might allow the patients to increase step length despite limitations in hip extension (Leigh et al., 2016; Thurston, 1985). Our subgroup analysis did not show a significant difference in peak hip extension angle for individuals with mild or moderate HOA. Thus, compensatory pelvic motion might not or only to a limited extent be necessary in this subject group. Yet, this has to be evaluated in future studies.

The results from our analyses show that modifications of kinematic patterns are not limited to the ipsilateral side or the affected joint but rather are a complex interplay of changes occurring at the pelvis and both lower limbs. These results are in line with those from whole-body analyses that show the highest discriminatory capacity in hip, knee and ankle sagittal angles and partly frontal plane ankle angles (Ardestani & Wimmer, 2016; Laroche et al., 2014; Ridgway et al., 2015).

4.5.2 Effects of hip osteoarthritis on stair walking kinematics

For stair ascent, no significant differences between HOA and CON participants were found during the meta-analysis. However, individual study results found decreased hip peak flexion during swing, causing a decrease in sagittal hip RoM. Stair ascent requires a greater sagittal hip RoM than level walking, and a high peak flexion is crucial for ensuring step clearance and avoidance of tripping (Meyer et al., 2015). Thus, decreased peak flexion and sagittal RoM might make individuals with HOA prone to falling during stair ascent. Likewise, peak abduction and external rotation of the hip were reduced, causing reduced hip RoM in the frontal and transverse planes. In contrast, the knee and ankle joints seem to be more externally rotated. Meyer et al. (2016) reported reduced peak adduction during stair ascent as a strategy for a wider base of support, which was not present in our meta-analysis. Thus, adopting a toe-out gait by externally rotating the foot and tibia might be another strategy to broaden the area of support for the stance limb (Cui et al., 2019).

For stair descent, the meta-analysis did not yield any significant differences in hip peak angles. However, an individual study still reported reduced sagittal hip RoM (Hall et al., 2017). Additionally, an increase in peak knee extension as well as peak plantar flexion was observed in one study (Popovic et al., 2021). This might stem from an attempt to prolong contact with the stair at TO or to contact the stair sooner to reduce single-support time. Yet, this is speculative and warrants further investigation.

As in stair ascent, a more pronounced external rotation of the knee and ankle joints was observed.

Overall, it has to be considered that only two studies analysed stair walking in individuals with HOA and thus, insights are still very limited. Additionally, the staircases used during the studies contained two or four steps, with the force plate positioned at the first step. Hence, the analysed step always contained movement initiation and/or transition to level walking. Results from Alcock et al. (2014) show large differences in hip, knee and ankle sagittal angles during steady-state stair ascent and the transition from gait to stair ascent. Hence, both of the retrieved studies give information on stair transition but not on steady-state stair walking.

4.5.3 Effects of hip osteoarthritis on turning kinematics

In their study on 45° turns during walking, Tateuchi et al. (2014) found reduced peak hip extension angles, similar to level walking. Additionally, peak hip flexion was also reduced. The step turn has been found to require more hip abduction than straight walking, while the spin turn requires greater hip adduction (M. Taylor et al., 2005). Individuals with HOA showed decreased peak abduction or adduction during these tasks, respectively (Tateuchi et al., 2014), but our meta-analysis did not yield these results for straight walking. Thus, analysis of turning while walking might be a beneficial extension of common gait analysis, especially as turning movements are encountered frequently during activities of daily living (Glaister et al., 2007). However, further studies on turning are needed to confirm these results and also to expand the knowledge on transverse plane kinematics, which are also more demanding during turning than straight walking (M. Taylor et al., 2005).

4.5.4 Limitations

Alongside the strengths of our study, there are also some limitations. As stated before, a lot of the calculations were based on a limited number of studies and thus have to be interpreted with caution. Yet, the results might still be helpful to identify research gaps and support hypothesis formulation for future studies. To have sufficient data for the meta-analysis, studies using different gait conditions (overground vs. treadmill) as well as different walking speeds and subject characteristics were included in the same analysis. Thus, the calculated SMD represents the average effect across a variety of study designs, measurement methods and subject populations. We therefore included prediction intervals in our analyses to provide a range for a potential HOA effect that might occur in individual study settings. Most of the significant results of our meta-analysis prediction intervals cross zero, meaning that in some settings, no difference between the HOA and CON groups might be present. However, it has to be borne in mind that prediction intervals may be overly wide when they are calculated from a limited number of studies or from studies at a higher risk of bias (Riley et al., 2011). Likewise, we included the I^2 statistic to estimate between-study heterogeneity. However, the I^2 value can be imprecise and biased (Hippel, 2015), especially in small meta-analyses such as the present one. Therefore, I^2 estimates have to be interpreted with caution. Lastly, we refrained from calculating a meta-regression to explore potential sources for between-study heterogeneity due to the small number of included studies (Borenstein, 2009, p. 221).

4.6 Conclusion

In summary, this was the first review to synthesise data on lower body joint kinematics during locomotion tasks in individuals with HOA. A total of 47 reports from 35 individual studies were retrieved.

Most studies focused on gait, where kinematic differences were found in the ipsi- and contralateral hip and knee joints. While changes at the hip occurred in all three motion planes, changes in knee kinematics occurred mainly in the sagittal plane. Differences were found between individuals with mild or moderate HOA and those with severe HOA. Thus, motion analysis for HOA patients should not exclusively focus on the kinematics of the affected hip joint but also include analysis of adjacent and contralateral joints. Despite no significant results of the meta-analysis in ankle or pelvis kinematics, several indications exist for further analyses in this area. Additionally, three studies on stair walking and turning while walking were reviewed, and both of these movement tasks might be promising extensions to clinical movement analysis due to their elevated requirements on joint mobility. Overall, large heterogeneity was observed across studies, so future studies have to further clarify the role of factors such as OA severity, laterality, age, gender, study design or movement execution in the analysis of lower limb joint kinematics.

5 Manuscript II - Test-retest reliability of hip proprioception assessment

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5.1 Abstract

Ageing and several illnesses, such as HOA, have been associated with impaired proprioception, leading to a decrease in motor control and a subsequent increase in the risk of falling and injury. Hence, there is a need for valid and reliable methods to assess joint proprioception in order to quantify impairments resulting from ageing or injury and the effectiveness of treatment strategies. Joint position reproduction (JPR) tests are a common and feasible method to assess joint proprioception. Studies analysing the reliability of hip joint JPR tests are sparse, and results are moderate at best. Therefore, the aim of this study was to assess the reliability of a refined hip flexion JPR test.

Inter-session reliability was assessed in two sessions (interim period of one week) in a population of 16 healthy young adults (age: 24.9 ± 2.2). Active-active hip flexion JPR was conducted in two angle ranges: 10-20° (LA) and 40-50° (HA). The blindfolded participants stood upright on one limb and performed hip flexion with a flexed knee joint (five repetitions per angle range). Target angles were set visually without an obstruction object. Positioning and repositioning events were defined autonomously by the subject by pressing a switch attached to a handlebar.

Hip flexion angles were analysed using a five-piece marker set (lateral knee, left & right anterior superior iliac spines (ASIS), left and right posterior superior iliac spines (PSIS)) captured by a motion capture system (200 Hz). Mean absolute errors (AE) between positioning and repositioning angles were calculated across five trials. Paired t-test, ICC (2,k), standard error

of measurement (SEM) and coefficient of variability (CV) were calculated to check for systematic bias as well as relative and absolute reliability.

AE for test and retest was $3.60 (\pm 1.42)$ and $3.14 (\pm 1.32)$ for LA and $2.88 (\pm 1.66)$ and $2.79 (\pm 1.28)$ for HA. No significant systematic error was observed between sessions (LA: $p = 0.29$; HA: $p = 0.80$). Intra-session ICC was 0.41 (95%CI: -0.62 0.79) for LA and 0.74 (95%CI: 0.24 0.91) for HA. SEM was 1.05° for LA and 0.75° for HA. The CV was 31.2 % for LA and 26.4 % for HA.

The results obtained in this study are in accordance with those reported by Benjaminse et al. (2009) (ICC 0.74, SEM: 0.72). They, however, used a constant target angle of 45° , which could have decreased variability. The resultant ICCs are larger than those presented by Arvin et al. (2015) (ICC: 0.11/0.56; SEM 0.6-0.7). Surprisingly, their described AE and inter-subject variability ($1.2\text{-}1.6 \pm 0.4\text{-}1.0$) were smaller than in the present study, despite analysing an older population (age: 73.5 ± 7.8) with potentially decreased proprioceptive capacity. In general, the JPR test in this study demonstrated moderate to fair reliability, with better results for larger flexion angles. Due to the small observed AE, combined with high inter-subject variability, the CV is relatively high. As a result, large group or treatment effects are required to detect statistically significant changes in joint proprioception.

5.2 Introduction

Precise afferent information about a human's position in and interaction with their environment is crucial for effective and accurate movement execution (Riemann & Lephart, 2002). Besides vestibular and visual information, humans rely on afferent internal information arising from sensory receptors in our skin, joints, tendons, and muscles (Proske & Gandevia, 2012). This information, in its entirety, is termed proprioception and includes multiple components such as knowledge of spatial position and joint angles (joint position sense (JPS)), knowledge of limb or joint movement (kinesthesia), as well as knowledge of mechanical load (force sense) (McCloskey, 1978).

Ageing as well as several illnesses and injuries, such as ruptures of the anterior cruciate ligamentum or OA, have been associated with impairments in proprioception (Pai et al., 1997; Relph et al., 2014; Wingert et al., 2014), leading to a decrease in motor control and subsequently an increase in the risk of falling and injury (Benjaminse et al., 2009). For individuals with HOA, a reduced number of sensory receptors (Moraes et al., 2011) and a loss of

vibratory sense have been described, leading to a decrease in joint position sense of the hip (Alkhamis et al., 2024). Furthermore, proprioceptive inaccuracy has been identified as a risk factor for functional decline in people with HOA (Dekker et al., 2009). Hence, enhancing joint proprioception is strived for through treatment (Beard et al., 1994; Birmingham et al., 2001a; Maitre et al., 2013; Patti et al., 2017; Rose, 2011; Stryła et al., 2013; Tsauo et al., 2008).

However, despite the substantial impact on joint stability, the alterations in proprioception due to illnesses or ageing observed in various proprioception tests are of rather small amplitude (X. Chen & Qu, 2018; Relph et al., 2014; Shakoor et al., 2014). Therefore, a precise assessment of joint proprioception is of high relevance for accurately quantifying the impairments experienced by patients or evaluating the effectiveness of treatment interventions.

There exists a wide range of methods to assess the different components of joint proprioception, of which the threshold for detection of passive movement as well as JPR tests have been most widely used (Hillier et al., 2015). A variety of JPR tests have been established, primarily differing in whether the joint angle position is adopted actively or passively during positioning and repositioning (e.g. active-active or passive-active) and whether repositioning is performed with the same or opposite limb (ipsilateral matching, contralateral matching). Thereby, previous research has demonstrated lower reproduction errors in active-active JPR test, probably due to the increase of afferent information arising from muscle spindles (Laufer et al., 2001). Additionally, it is suggested that testing in an upright, weight-bearing situation is more representative of joint positions that occur during functional activities and sports. Thus, active-active JPR test in weight-bearing situations might possess an increased ecological validity compared to passive JPR tests or tests conducted in sitting or supine positions (Stillman & McMeeken, 2001). For the knee joint, several testing procedures have been established and verified regarding the quality criteria (Arvin et al., 2015; Piriyaprasarth et al., 2008; Smith et al., 2013; Stillman & McMeeken, 2001; Suner-Keklik et al., 2017).

In contrast, studies focusing on the reliability of JPS assessment at the hip joint are sparse. Benjaminse et al. (2009) reported an intra-session ICC of 0.23 and an inter-session ICC of 0.74 for an active-active JPR test with constant hip flexion angles of 45° in a population of healthy young adults. The study conducted by Arvin et al. (2015) reported intra-session ICCs of 0.11 and 0.56 for the left and right legs, respectively, in a healthy elderly population performing a JPR test with variable hip flexion angles between 10 and 45°. A passive-active JPR measurement yielded inter-session ICCs of 0.67 to 0.73 in healthy participants (Shakoor et al., 2014). The modest reliability and high variability observed between studies and limbs might be explained by several methodological limitations in the conception of the testing proce-

ture. Benjaminse et al. (2009) stated that subject concentration might have been impaired due to problems in maintaining balance in a blindfolded single-stance situation. Additionally, learning effects might have occurred due to constant target angles. Furthermore, additional sensory feedback might arise from the usage of an obstructive object to define the target angle during limb positioning. In the study by Arvin et al. (2015), hip flexion movement was performed with an extended knee joint, which might restrict hip flexion mobility and generate additional afferent information due to strain on the biarticular hamstring muscles. Secondly, external triggering of events (positioning, repositioning) by the test operator, as applied by Arvin et al. (2015), may cause a time delay and thus a discrepancy between the subject's perceived position and the recorded hip angle.

Therefore, the aim of this study was to refine the testing procedure of an active-active hip flexion angle reproduction test and subsequently to assess the inter-session test-retest reliability of this procedure.

5.3 Materials and Methods

5.3.1 Participants

A total of 16 healthy young volunteers participated in this study (age: 24.9 ± 2.4 years; height: 177.1 ± 7.5 cm; weight: 72.2 ± 10.7 kg; BMI: 22.9 ± 2.1 kg/m²; 6 females). All participants provided written informed consent prior to participating in the study. The study was approved by the institutional review board.

5.3.2 Study procedure and data acquisition

The reliability of a hip flexion JPR test was evaluated using an inter-session intra-rater test-retest design. Participants were tested twice at the same time of day, with an interim period of one week, by the same test operator. Two different target angle ranges were presented to the participants: Hip flexion angles of 10-20°, termed low angle (LA) and hip flexion angles of 40-50°, termed high angle (HA). The LA range was used to account for the difference in proprioception at the end ranges of motion (Borsa et al., 1997), whereas the angles of the HA condition represent peak angles encountered during activities of daily living (e.g. walking, stair descent) (Riener et al., 2002). Participants were tested alternately using either the left leg or the right leg (right $n = 8$; left $n = 8$) based on their order of inclusion in the initial test-

ing session, with the same limb used consistently in both sessions. Motor limb dominance was assessed using a ball kicking task (dominant $n = 9$; non-dominant $n = 7$) (Chapman et al., 1987; Schneiders et al., 2010).

The procedure of the angle reproduction test was designed based on the testing procedure described by Arvin et al. (2015), yet including some structural innovations aiming to increase test reliability.

Firstly, participants were equipped with five reflective markers positioned directly onto the skin at the left and right ASIS and PSIS and the lateral epicondyle of the ipsilateral knee. 3D marker trajectories were recorded using a 12-camera motion capture system (200 Hz; Vicon Motion Systems, Oxford Metrics Group, Oxford, UK). Participants then received a standardised instruction, including a demonstration of the test task, followed by two practice trials.

Participants were positioned on a block of 20 cm height in an upright standing position. The contralateral foot was placed along the edge of the block, allowing the ipsilateral lateral leg to swing freely. Participants were blindfolded to eliminate any visual information and were barefoot to exclude any effects of footwear. To avoid loss of balance and assure that the focus of attention is the perception of joint position, participants hold onto a horizontal bar positioned laterally on their contralateral side. The ipsilateral arm was held across the chest (Figure 5.1). The “Start” command was given, and participants started to actively flex their hip joint in a slow and steady manner. In contrast to the study by Arvin et al. (2015), participants were allowed to flex their knee joint to avoid strain on the hamstring muscles. When the participant had reached the target angle area, an acoustic “Stop” signal was given, and participants stopped the movement and memorised the angle position. When the participants had sufficiently memorised the joint position, they pressed a button, positioned at the horizontal handlebar, which activated an infrared light visible in the simultaneous motion capture recordings. This is in contrast to the test procedure of Arvin et al. (2015), where events were triggered by the test operator after verbal command. With the command “OK”, participants were allowed to return to the initial position where they rested for three seconds. Afterwards, another “Start” command was given and participants replicated the angle as precisely as possible, pushing the switch button when they felt to be in the correct position. With the command “OK”, participants again returned to the initial position and rested for 30 seconds before the next trial. In this period, participants were allowed to place both feet on the support box to avoid muscle fatigue. Five trials for each angle range were recorded in a randomised order, resulting in a total of ten trials.



Figure 5.1: Measurement setup for joint position reproduction test.
Note. Box height = 20 cm; handlebar with attached switch.

5.3.3 Data processing and analysis

Data were processed using Vicon Nexus software (version 1.8.5) and MATLAB (version R2017b; The MathWorks Inc., Natick, MA, USA). Marker trajectories were filtered using a 2nd order Butterworth low-pass filter with a cut-off frequency of 10 Hz. The hip joint centre (HJC) was calculated using the formula proposed by Harrington et al. (2007). Sagittal plane hip flexion angle was calculated between the pelvis segment (vector between the midpoint of the ipsilateral ASIS and PSIS markers and HJC) and the thigh segment (vector between the lateral knee marker and HJC). The hip angle obtained in the initial position (hip fully extended) was defined as 0, and the offset was subtracted from all joint angles. Absolute errors between the prescribed and the replicated hip angles were calculated for each trial and averaged across the five trials of each angle range for further analysis.

5.3.4 Statistical analysis

All statistical tests were performed using IBM SPSS Statistics (version 25.0). Data from the two target angle ranges (LA and HA) were analysed separately for reliability. Prior to further analysis, all data were analysed for normal distribution using a Shapiro-Wilks normality test. Systematic error between test sessions and differences in the prescribed angles were analysed using dependent sample t-tests. To assess differences in movement velocity between sessions as well as between positioning and repositioning, a two-way (session, event) ANOVA was used. The level of significance was set *a priori* to $\alpha < 0.05$.

A two-way mixed effects, absolute agreement model of intra-class correlation coefficient (ICC 2,k) (Shrout & Fleiss, 1979) was employed to investigate the relative inter-session intra-rater test-retest reliability. Correlation coefficients were interpreted according to Landis & Koch (1977), with values of 0.2-0.4 considered fair, 0.4-0.6 moderate, 0.6-0.8 substantial and 0.8-1.0 almost perfect reliability.

To analyse absolute reliability, the SEM and CV were calculated as follows (Atkinson & Nevill, 1998; Thomas & Nelson, 1990):

$$SEM = SD * \sqrt{(1 - ICC)}$$

$$CV = SEM / Mean$$

Based on the SEM, the minimal detectable change (MDC) was calculated as:

$$MDC = SEM * 1.96 * \sqrt{2}$$

5.4 Results

For each subject, mean absolute errors and SD of the five trials of each session and angle range are presented in Table 5.1.

Analyses of the prescribed target angle range can be found in Table 5.2. Prescribed angles did not differ between sessions. However, the mean prescribed angles (38.6 and 39.4°) for the HA condition were outside of the strived angle range of 40-50°.

Movement velocity increased from session one to session two in both target angle ranges, but did not differ between positioning and repositioning (Table 5.3).

Moderate ICC was found for the LA condition, while a substantial ICC was found for the HA condition. No change in AE was found between sessions. Overall, larger AE were observed in the LA than in the HA angle range, while SEM, CV and MDC were larger than those observed for HA Table 5.4.

Table 5.1: Descriptive statistics of the absolute error (AE) for each subject, session and angle range.

Subject ID	High angle		Low angle	
	AE Test [°]	AE Retest [°]	AE Test [°]	AE Retest [°]
	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)
1	1.87 (0.65)	0.90 (0.70)	2.97 (2.12)	4.71 (1.27)
2	2.39 (1.46)	2.71 (2.12)	4.44 (2.69)	1.42 (1.04)
3	2.09 (1.39)	2.54 (0.65)	2.09 (1.95)	0.87 (0.83)
4	1.91 (0.79)	1.02 (0.67)	1.82 (0.41)	1.59 (1.45)
5	6.97 (1.78)	3.77 (2.34)	5.72 (3.41)	3.65 (2.21)
6	6.31 (1.87)	4.34 (1.69)	4.91 (1.71)	4.94 (2.28)
7	2.22 (0.78)	3.15 (1.98)	1.79 (1.70)	3.50 (1.08)
8	1.70 (0.86)	3.17 (2.39)	2.61 (1.51)	4.79 (1.73)
9	2.60 (1.46)	1.05 (0.90)	2.49 (2.29)	1.93 (1.52)
10	1.69 (1.33)	1.11 (1.01)	2.68 (1.55)	2.35 (0.95)
11	4.02 (3.08)	4.62 (2.76)	3.48 (1.85)	4.96 (1.01)
12	1.84 (1.18)	2.01 (1.02)	3.80 (1.90)	3.48 (1.84)
13	4.10 (1.93)	3.86 (2.02)	2.84 (1.33)	2.22 (1.73)
14	2.78 (1.52)	4.20 (2.34)	4.50 (0.94)	3.90 (1.74)
15	2.13 (1.01)	3.85 (2.35)	5.24 (2.85)	3.10 (0.77)
16	1.45 (0.90)	2.33 (1.52)	6.29 (4.05)	2.88 (2.04)

Table 5.2: Descriptive and inferential statistics of the prescribed angles for each session and angle range.

Angle range	Prescribed Angle Test [°]	Prescribed Angle Retest [°]	p
	Mean (SD)	Mean (SD)	(t-test)
Low (LA)	18.8 (5.3)	19.6 (4.8)	0.71
High (HA)	38.6 (5.6)	39.4 (4.4)	0.38

Note. P-values as revealed by dependent sample t-test.

Table 5.3: Descriptive and inferential statistics of the movement velocity for each event, session and angle range.

Angle range	Angle Vel. E1 Test [°/s]	Angle Vel. E1 Retest [°/s]	Angle Vel. E2 Test [°/s]	Angle Vel E2 Retest [°/s]	Session	Event	Interaction
	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	p	p	p
Low (LA)	14.0 (4.2)	16.5 (5.4)	12.0 (4.5)	15.2 (4.9)	< 0.01	0.21	0.33
High (HA)	11.9 (4.0)	19.3 (8.3)	9.9 (4.3)	17.2 (6.5)	< 0.01	0.14	0.96

Note. P-values as revealed by two-way repeated measure ANOVA; significant results ($p < 0.05$) in bold. Vel. = velocity; E1 = positioning; E2 = repositioning.

Table 5.4: Descriptive and inferential statistics of the absolute error (AE) for each session and angle range.

Angle range	AE Test [°]	AE Retest [°]	ICC	SEM [°]	CV [%]	MDC [°]	p
	Mean (SD)	Mean (SD)	(95%CI)				t-test
Low (LA)	3.60 (1.42)	3.14 (1.32)	0.41 (-0.62 0.79)	1.05	31.23	4.12	0.29
High (HA)	2.88 (1.66)	2.79 (1.28)	0.74 (0.24 0.91)	0.75	26.44	2.08	0.80

Note. P-values as revealed by dependent sample t-test. ICC = intraclass correlation coefficient; 95%CI = 95 % confidence interval; SEM = standard error of measurement; CV = coefficient of variation; MDC = minimal detectable change.

5.5 Discussion and Conclusion

The aim of this study was to assess the inter-session test-retest reliability of a modified joint position testing procedure for the assessment of hip proprioception in the sagittal plane.

Based on the correlation analysis, the testing procedure is of moderate reliability for the LA range and substantial reliability for the HA range. No systematic bias was observed between testing sessions.

The observed results for HA, despite several methodological differences, are in accordance with those reported by Benjaminse et al. (2009) (AE: 2.8°; ICC 0.74; SEM: 0.72). They, however, used a constant target angle of 45°, which increases the risk for learning effects and thus, systematic error. Yet, as opposed to the present study, systematic error was not assessed and the applied ICC (3,k) analyses for consistency and not agreement (Shrout & Fleiss, 1979)

The resultant ICCs observed in the present study are larger than those presented by Arvin et al. (ICC: 0.11-0.56; SEM 0.6-0.7), indicating that refinements of the testing procedure as applied in the current study were able to increase reliability. Surprisingly, their described AE (1.2-1.6) were smaller than in the present study, despite analysing an older population (mean age: 73.5 ± 7.8) with potentially decreased proprioceptive capacity (Wingert et al., 2014). Conducting hip flexion with an extended knee joint, as done in the study by Arvin et al. (2015), causes strain on the biarticular hamstring muscles. This additional source of proprioceptive information could decrease the positioning error during the JPR task; however, it is not based on afferent information arising from the hip joint and its surrounding structures.

Due to high interindividual variability, the SEM and CV observed in this study are relatively high, leading to an MDC of about 4° for the LA and 2° for the HA range. Thus, effects of illness or treatment have to be sufficiently large to demonstrate statistically significant differences, and differences are more likely to be found when testing is conducted in the HA range.

While using an active-active JPR test in an upright standing position best reflects functional use of proprioception (Hillier et al., 2015), it requires sufficient kinesthetic memory, muscle strength and motor control. Performing the JPR test in a supine position might reduce variability and thus increase reliability, however, at the cost of ecological validity.

In summary, the JPR test in this study demonstrated moderate to substantial reliability, with better results for larger hip flexion angles. Due to large inter-subject variability, the SEM and CV are relatively high, raising the need for large group or treatment effects to detect significant changes in joint proprioception.

6 Manuscript III - Hip mobility requirements and joint loading during straight walking and turning

Slightly modified version of the manuscript published as:

Steingrebe, H., Sell, S., & Stein, T. (2024). Mobility Requirements and Joint Loading during Straight Walking and 90° Turns in Healthy Older People and Those with Hip Osteoarthritis. *Journal of Clinical Medicine*, 13(17), 5021. <https://doi.org/10.3390/jcm13175021>

6.1 Abstract

Hip mobility and joint loading in HOA patients are mostly assessed during straight walking. Yet, mobility limitations in the frontal and transverse planes are rarely found during this task in individuals with mild-to-moderate symptoms. Turning movements are frequently encountered during everyday life and might require larger hip mobility than straight walking, especially in the frontal and transverse planes. Thus, hip mobility and hip loading during straight walking and 90° turns in persons with HOA and healthy older adults were compared in this study.

A retrospective analysis was conducted on 21 individuals with mild-to-moderate HOA and 21 healthy controls. Hip angles and moments were assessed during straight walking and 90° step and spin turns. Gait analysis was conducted using a motion capture system and a force plate. Group and movement task differences were assessed with a mixed-model ANOVA.

Peak abduction and adduction angles were largest during the step and spin turn, respectively, as were the group differences between HOA and healthy participants. Both turns require greater transverse hip RoM than straight walking. Limitations in transverse hip mobility in the HOA group were especially prominent during the step turn. Both turns cause higher joint moments than straight walking.

The additional inclusion of 90° step and spin turns in gait analysis can enhance early identification of hip mobility limitations in the frontal and transverse planes in individuals with mild-to-moderate HOA. Early diagnosis is crucial for the timely application of conservative treatment strategies.

6.2 Introduction

Patients with HOA suffer from pain and loss of function, leading to a decrease in quality of life (Salaffi et al., 2005). Functional loss is mainly caused by muscle weakness (Loureiro et al., 2018) and limitations in passive and active hip RoM (Baker et al., 2016). Previous analyses of active hip RoM have mainly focused on straight walking (Constantinou et al., 2017; Leigh et al., 2016; Steingrebe et al., 2023). Likewise, joint moments, as a surrogate measure of joint loading, are also mostly assessed during straight walking. In a meta-analysis on HOA-induced changes in peak hip flexion and adduction moments during walking, differences were found for individuals with severe HOA but not for those with only mild-to-moderate symptoms (Diamond et al., 2018). However, joint mobility requirements and joint loading in terms of external moments might be larger during other movements of daily living, especially in the frontal and transverse planes.

One task that has gained attention from various research fields is turning or curve walking (Courtine & Schieppati, 2003; Dixon et al., 2013; Hicheur et al., 2005; Krafft et al., 2015). Turning movements are encountered frequently during activities of daily living, as 35-45 % of daily steps are turning or nonlinear steps (Glaister et al., 2007) and turns are a risk factor for falls (Leach et al., 2018). Turning movements can be conducted either in a spin or step turn manner (Hase & Stein, 1999). Both movement patterns require more pronounced external rotation of the hip, step turning requires larger hip abduction, and spin turning requires larger hip adduction than straight walking (M. Taylor et al., 2005). Yet, these biomechanical insights have been obtained in a sample of healthy young adults. In HOA patients, limited passive internal rotation and adduction are often core criteria for functional HOA diagnosis (Katz et al., 2021) and limitations in external rotation mobility have been associated with higher levels of disability (Steultjens et al., 2000). To the best of our knowledge, only one study of people with HOA included turning movements. This study showed decreased peak hip abduction and adduction as well as peak hip adduction moments during 45° turns compared to age-matched healthy controls (Tateuchi et al., 2014). However, knowledge of transverse plane biomechanics is still lacking, and RoM requirements and joint moments of different movement tasks have not been compared. Additionally, it has been found that the

majority of turns during daily life are between 76° and 120° (Sedgman et al., 1994), meaning that turns in this range presumably are of higher practical relevance.

Therefore, this study aimed to compare hip mobility, in terms of RoM and peak angles, and joint loading in terms of peak external hip moments, between straight walking and 90° step and spin turns in healthy older adults and in individuals with mild-to-moderate HOA. We hypothesised that 90° turns would require larger peak hip angles and overall hip RoM, and would provoke higher joint moments in the frontal and transverse planes than straight walking. Subsequently, we hypothesised that mobility deficits as well as modifications in joint loading of individuals with HOA would be more pronounced during turning than during straight walking.

6.3 Materials and Methods

In this secondary analysis, data from a study on the effects of hip bracing in different movement tasks were reanalysed. Therefore, data have been published in part elsewhere (Steingrebe et al., 2022).

6.3.1 Participants

Participants for this study were recruited from local physiotherapy practices, via local newspapers, or from the university community and thus, represent a sample of convenience. In total, 42 individuals participated in this study, thereof 21 individuals with mild-to-moderate unilateral primary HOA (10 females, age: 64.0 ± 9.6 years; BMI: 24.2 ± 2.9 ; Harris Hip Score (HHS): 74.6 ± 11.8); and 21 healthy participants (10 females, age: 63.1 ± 9.2 years; BMI 25.2 ± 2.7 ; HHS: 98.4 ± 2.3). Groups did not differ in terms of age, height, body mass, or BMI. Additionally, participants from the control group were randomly and counterbalanced assigned to a left or right group to match the total number of affected right and left limbs of the HOA group. Participants in the HOA group had radiologically confirmed HOA of K-L grade two or higher, hip pain during activities of daily living and decreased hip function (HHS ≤ 95). The healthy control participants had no radiological signs of HOA (K-L Score ≤ 1), did not experience hip pain during activities of daily living and had good hip function (Harris Hip Score > 95). Detailed inclusion and exclusion criteria can be found in Table 6.1. The study procedure was approved by the ethics committee of the Karlsruhe Institute of Technology (KIT). All participants provided written informed consent prior to study participation.

Table 6.1: Inclusion and exclusion criteria

Group	Inclusion criteria	Operationalisation
HOA	Radiologically confirmed HOA	K-L-Score of 2-4
	Symptomatic HOA	Hip pain during activities of daily living within the last 3 months
	Decreased hip function	Harris Hip Score between 65-95
	Unilateral HOA	Contralateral K-L-Score ≤ 2 ; Contralateral hip pain free within the last 3 month; Unrestricted passive RoM of contralateral hip (sagittal RoM $\geq 90^\circ$; transverse RoM $\geq 15^\circ$; peak abduction $\geq 20^\circ$; flexing contracture $\leq 10^\circ$)
Healthy	No radiological signs of HOA	Bilateral K-L-Score ≤ 1 (if radiographic images available)
	No symptoms of HOA	No hip pain within the last 3 months during activities of daily living
	Good hip function	Harris Hip Score ≥ 96
Group	Exclusion criteria	
HOA	Secondary HOA caused by trauma	
	Contraindications of X-ray imaging	
Both	Body Mass Index $\geq 35 \text{ kg/m}^2$	
	Orthopaedic injury of other joints of the lower limbs and back (e.g. pain, osteoarthritis > grade 1 (self-reported), endoprosthesis, rheumatoid arthritis, acute herniated disc, etc.)	
	Neuromuscular disorders or neurological complaints (e.g. vertigo)	

Note. HOA = hip osteoarthritis; RoM = range of motion; K-L = Kellgren-Lawrence.

6.3.2 Testing protocol

Participants performed straight walking and pre-planned 90° step and spin turns (Figure 6.1) at a self-selected speed. 90° turns were chosen due to their high practical relevance as right-angle turns are frequently encountered in urban settings (Shi et al., 2022) and the widespread usage of this turn angle in previous studies (Dixon et al., 2018; Nolasco et al., 2019; M. Taylor et al., 2005) as well as their inclusion in functional tests (Deathe & Miller, 2005). Movement velocity was controlled using light barriers and kept constant (within $\pm 5\%$ of the first trial) across trials of the same movement task. For each movement task, five valid trials were recorded. During the turning trials, participants initiated the turn within a marked corridor of 50 cm on the force plate. To standardise movement execution, participants were instructed to perform the turn in an abrupt manner, resulting in an entire 90° turn within one GC (Xu et al., 2004). Turning was always conducted with the affected limb for the HOA group and a matched limb for the control group (Figure 6.1).

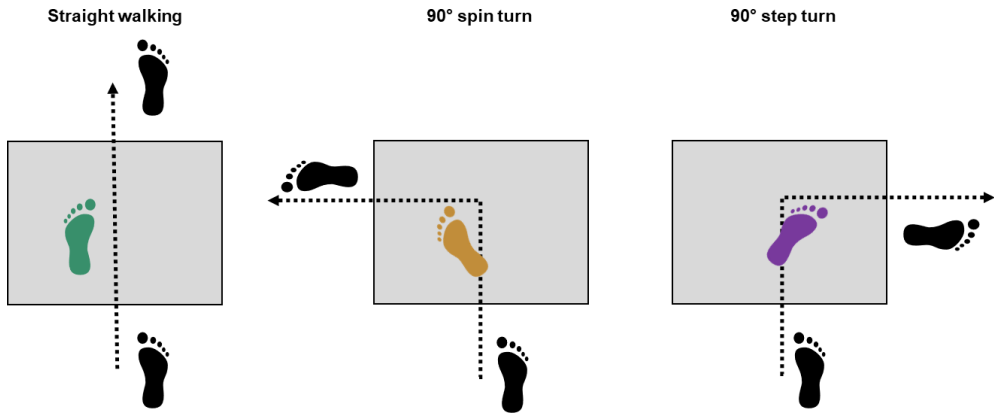


Figure 6.1: Schematic representation of foot placement during straight walking, 90° spin turn and 90° step turn. *Note.* Coloured feet = affected limbs in hip osteoarthritis participants / matched limbs in healthy controls. Green = straight walking, purple = 90° step turn, yellow = 90° spin turn.

6.3.3 Gait analysis and data processing

Whole-body movement was assessed with a 16-camera motion-capturing system (200 Hz; Vicon Motion Systems, Oxford Metrics Group, Oxford, United Kingdom). Simultaneously, GRF were recorded using a 3D force plate (1,000 Hz; Advanced Mechanical Technology Inc., Watertown, MA, United States). Kinematic and GRF data were filtered using a 4th-order Butterworth low-pass filter with a cut-off frequency of 15 Hz. A full-body marker set with 42 retroreflective markers was applied (Härtel & Hermsdorf, 2006). 3D hip joint angles and external joint moments were calculated using an inverse kinematics and dynamics approach with the multi-body ALASKA Dynamicus model (Härtel & Hermsdorf, 2006), including the HJC definition proposed by Harrington et al. (2007). Peak hip joint angles and moments and the hip RoM during ST of straight walking or the turning step of 90° turns were assessed. Additionally, forward-oriented centre of mass (CoM) velocity at IC and TO, as well as ST duration, were analysed. Mean values across the five trials of each movement were calculated for each subject. For five participants, one trial of step or spin turning was detected as invalid after the measurement was taken. These trials were excluded from the analysis, and the mean value was calculated across the remaining four trials.

6.3.4 Statistical analysis

As the basis for this manuscript is secondary data, the *a priori* calculation of the sample size was based on effects induced by hip bracing during level walking in individuals with HOA (Nérot & Nicholls, 2017). Calculations were conducted using G*Power (version 3.1.9.3) (Faul et al., 2007). With observed effect sizes of 0.92 and 1.17 for peak hip adduction and internal rotation (Nérot & Nicholls, 2017), a significance criterion of $\alpha = 0.05$, and power of 0.95, the minimum sample sizes required were 18 and 12, respectively.

All statistical analyses were conducted using R (version 4.2.2) (R Core Team, 2023). Normal distribution and homogeneity of variance were tested using the Shapiro-Wilk and Levene tests, respectively. Statistical differences were assessed using a mixed-model ANOVA with movement task (straight walking, step turn, spin turn) as the within-subject factor, and group (healthy, HOA) as the between-subject factor. A Mauchly test was used to assess sphericity. If sphericity was violated, Greenhouse-Geisser estimates were used. Post hoc analyses were conducted using t-tests for dependent samples with Bonferroni corrections. The level of significance was set *a priori* to $\alpha < 0.05$. Partial eta squared (η_p^2) was used for effect size estimation.

6.4 Results

Means and SD of spatiotemporal parameters are presented in Table 6.2, kinematic parameters are presented in Table 6.3, and kinetic parameters are presented in Table 6.4. Additionally, hip joint mobility and joint dynamics are visualised in Figure 6.2. Hip joint angle and moment time curves in the sagittal, frontal and transverse planes are presented in Figure 6.3. Results from all statistical analyses can be found in Table A.2 and Table A.3 in the Appendix.

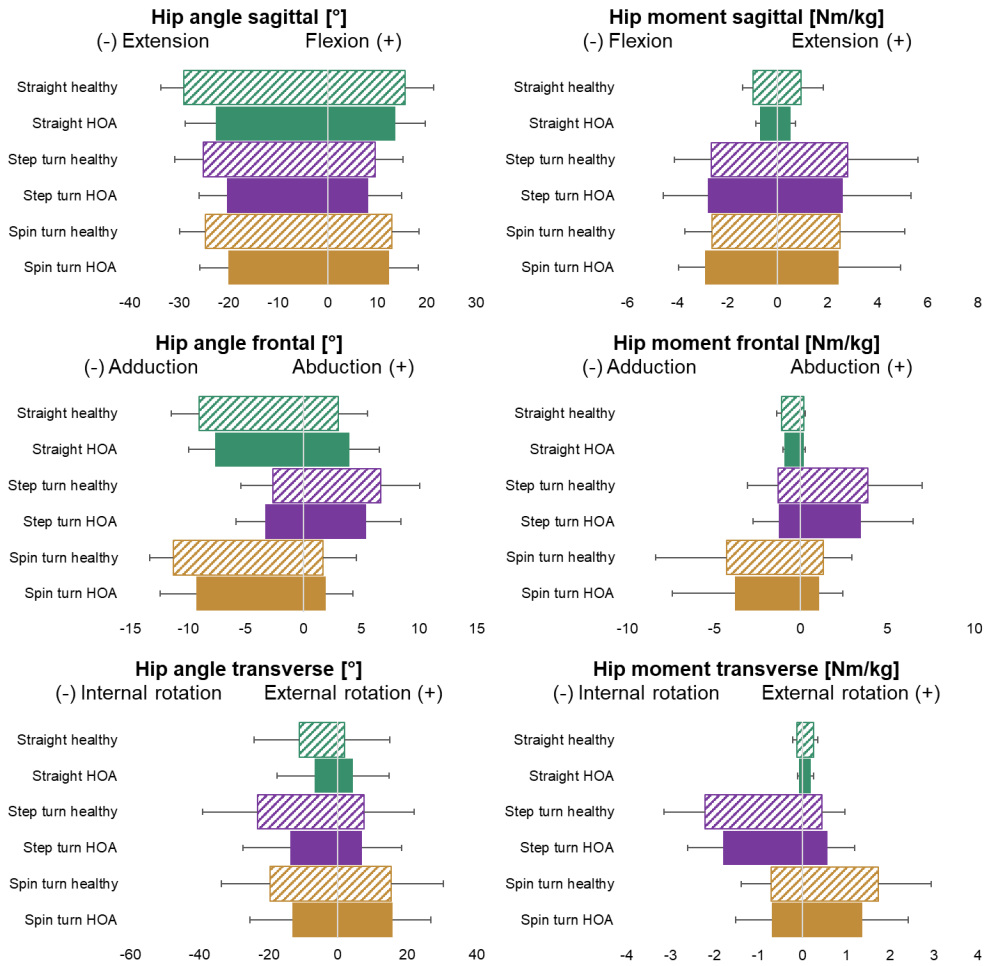


Figure 6.2: Peak hip angles [°] and external hip joint moments [Nm/kg].

Note. Data presented as mean $\pm$ SD. Green = straight walking, purple = 90° step turn, yellow = 90° spin turn. Solid bars = individuals with hip osteoarthritis, hatched bars = healthy controls.

Table 6.2: Descriptive results of the evaluated spatiotemporal parameters.

Parameter	Straight walking		90° step turn		90° spin turn	
	Healthy	HOA	Healthy	HOA	Healthy	HOA
	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)
CoM velocity IC [m/s]	1.36 (0.12)	1.27 (0.19)	0.97 (0.12)	0.94 (0.14)	0.96 (0.12)	0.92 (0.19)
CoM velocity TO [m/s]	1.39 (0.12)	1.29 (0.19)	0.98 (0.15)	0.94 (0.16)	0.97 (0.11)	0.93 (0.12)
Stance phase duration [s]	0.64 (0.05)	0.65 (0.05)	0.70 (0.08)	0.73 (0.10)	0.74 (0.08)	0.78 (0.14)

Note. HOA = hip osteoarthritis; IC = initial contact; TO = toe-off; CoM = centre of mass.

Table 6.3: Descriptive results of the evaluated kinematic parameters at the hip joint during the stance phase.

Parameter [°]	Straight walking		90° step turn		90° spin turn	
	Healthy	HOA	Healthy	HOA	Healthy	HOA
	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)
Peak flexion	15.70 (5.75)	13.69 (6.14)	9.64 (5.58)	8.18 (6.84)	12.95 (5.55)	12.49 (5.82)
Peak extension	-29.11 (4.5)	-22.56 (6.28)	-25.13 (5.76)	-20.32 (5.73)	-24.72 (5.25)	-20.08 (5.82)
Sagittal RoM	44.81 (5.52)	36.25 (8.34)	34.77 (5.58)	28.50 (8.01)	37.66 (5.25)	32.57 (6.49)
Peak abduction	3.02 (2.58)	4.02 (2.56)	6.71 (3.37)	5.47 (2.98)	1.71 (2.92)	1.95 (2.38)
Peak adduction	-9.04 (2.40)	-7.66 (2.30)	-2.61 (2.76)	-3.26 (2.53)	-11.25 (2.03)	-9.26 (3.12)
Frontal RoM	12.06 (1.60)	11.68 (2.69)	9.32 (2.41)	8.73 (2.39)	12.97 (1.81)	11.21 (3.00)
Peak ext. rotation	1.90 (13.2)	4.34 (10.5)	7.68 (14.3)	6.97 (11.5)	15.41 (15.0)	15.82 (11.1)
Peak int. rotation	-11.19 (13.1)	-6.59 (11.0)	-23.07 (16.0)	-13.82 (13.7)	-19.62 (13.9)	-13.20 (12.2)
Transverse RoM	13.08 (3.56)	10.93 (2.82)	30.75 (5.97)	20.79 (7.85)	35.03 (8.45)	29.02 (8.72)

Note. HOA = hip osteoarthritis; RoM = range of motion: ext = external; int = internal.

Table 6.4: Descriptive results of the evaluated external joint moments at the hip joint during the stance phase.

Parameter	Straight walking		90° step turn		90° spin turn	
	Healthy	HOA	Healthy	HOA	Healthy	HOA
	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)
Peak extension moment [Nm/kg]	0.93 (0.90)	0.53 (0.19)	2.82 (2.80)	2.61 (2.74)	2.50 (2.58)	2.45 (2.46)
Peak flexion moment [Nm/kg]	-0.96 (0.43)	-0.69 (0.18)	-2.63 (1.48)	-2.77 (1.80)	-2.62 (1.06)	-2.89 (1.06)
Peak abduction moment [Nm/kg]	0.20 (0.08)	0.19 (0.09)	3.89 (3.13)	3.50 (3.01)	1.33 (1.62)	1.09 (1.35)
Peak adduction moment [Nm/kg]	-1.10 (0.25)	-0.92 (0.10)	-1.29 (1.75)	-1.25 (1.47)	-4.28 (4.10)	-3.78 (3.63)
Peak ext. rotation moment [Nm/kg]	0.26 (0.09)	0.20 (0.06)	0.46 (0.52)	0.58 (0.61)	1.74 (1.22)	1.37 (1.06)
Peak int. rotation moment [Nm/kg]	-0.12 (0.11)	-0.07 (0.04)	-2.21 (0.94)	-1.81 (0.81)	-0.71 (0.69)	-0.69 (0.83)

Note. HOA = hip osteoarthritis; ext. = external; int. = internal.

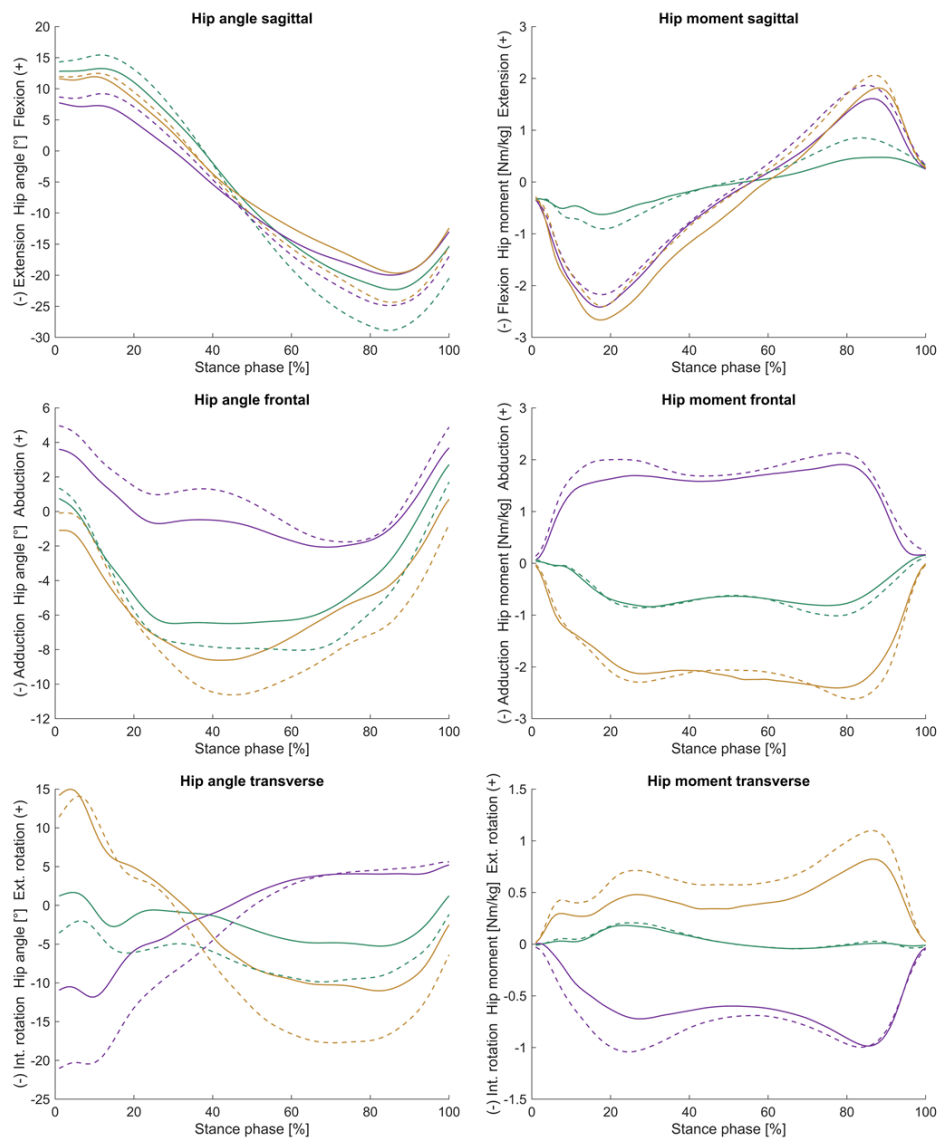


Figure 6.3: Sagittal, frontal and transverse hip joint angle [°] and external hip joint moment [Nm/kg] time curves across the stance phase [%].

Note. Data presented as mean. Green = straight walking, purple = 90° step turn, yellow = 90° spin turn. Dashed line = healthy controls; solid line = individuals with hip osteoarthritis.

6.4.1 Mobility requirements of the movement tasks

For all kinematic parameters, namely peak hip angles and hip RoM, a significant main effect for movement task ($p < 0.001$; η_p^2 0.39 - 0.91) was found. Post hoc comparisons showed that straight walking and step turn differed significantly ($p < 0.001$) for all kinematic parameters, with lower sagittal hip mobility and hip adduction but larger hip abduction and transverse mobility during the step turn. Likewise, straight walking and spin turn differed significantly in all kinematic parameters ($p < 0.001$) except for frontal hip RoM ($p = 1.00$), with lower sagittal hip mobility and hip abduction, but larger hip adduction and transverse mobility. Step and spin turn differed significantly ($p < 0.001$) for all kinematic parameters except for peak hip extension ($p = 1.00$) and peak hip internal rotation angle ($p = 0.16$). Thereby, lower hip flexion and sagittal hip RoM, hip adduction, and frontal hip RoM, as well as lower external rotation and transverse hip RoM, but larger hip abduction were found for the step turn compared to the spin turn.

6.4.2 Joint loading during the movement tasks

For all kinetic parameters, namely peak external hip joint moments, a significant main effect for movement task ($p \leq 0.006$; η_p^2 0.16- 0.78) was found. Pairwise comparisons showed that straight walking and step turn differed significantly for all kinetic parameters ($p \leq 0.004$) except for peak hip adduction moment ($p = 0.91$), with higher hip joint moments during the step turn. Likewise, straight walking and spin turn differed significantly in all parameters ($p < 0.001$), with higher joint moments during the spin turn. Step and spin turn differed significantly ($p < 0.001$) in frontal and transverse peak joint moments, with the step turn provoking higher peak hip abduction and internal rotation moments but lower hip adduction and external rotation moments.

6.4.3 Effect of hip osteoarthritis on hip mobility and joint loading

A significant main effect for group was found for peak hip extension angle ($p = 0.002$; $\eta_p^2 = 0.22$), sagittal hip RoM ($p = 0.001$; $\eta_p^2 = 0.24$) and transverse hip RoM ($p < 0.001$; $\eta_p^2 = 0.28$) with lower values in the HOA group. Significant interaction effects were found for peak hip adduction angle ($p < 0.001$), peak hip abduction angle ($p = 0.001$), peak hip internal rotation angle ($p = 0.046$), sagittal hip RoM ($p = 0.041$) and transverse hip RoM ($p = 0.006$) (Figure 6.2).

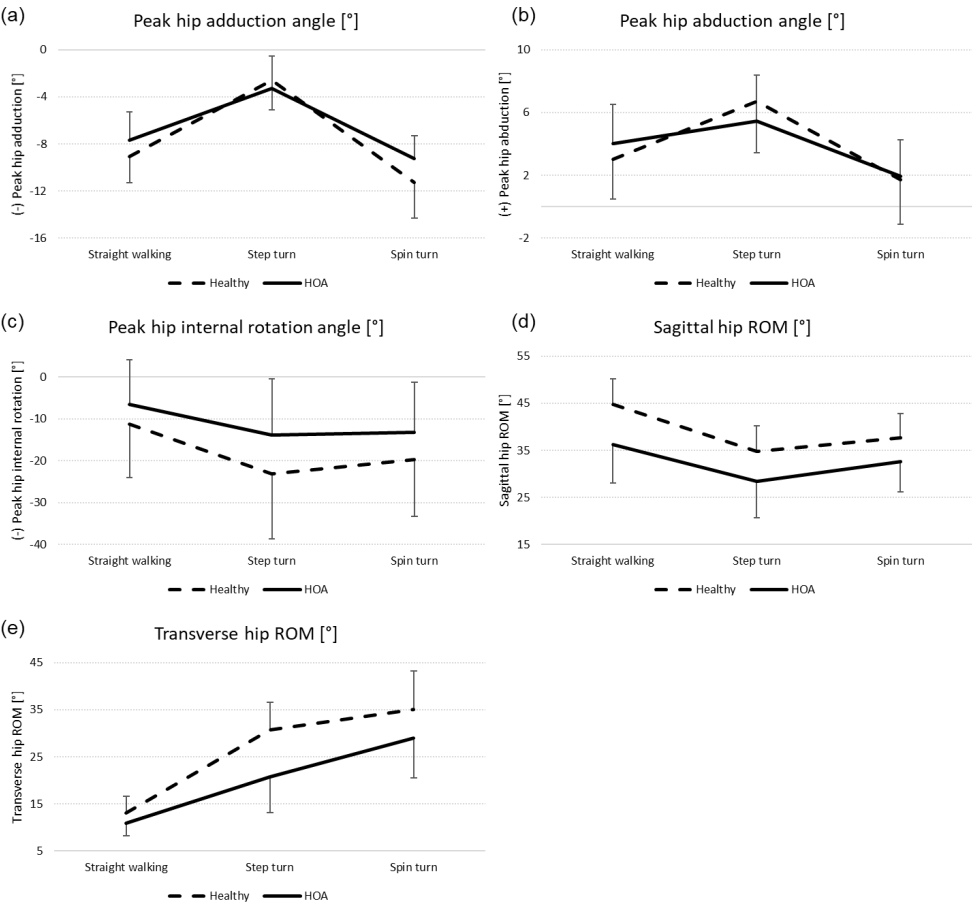


Figure 6.4: Parameters with significant interaction effect of group and movement task: (a) peak hip adduction angle, (b) peak hip abduction angle, (c) peak hip internal rotation angle, (d) sagittal hip range of motion (RoM), (e) transverse hip RoM. Dashed line = healthy controls; solid line = individuals with hip osteoarthritis.

Note. Data presented as mean $\pm$ SD.

6.5 Discussion

To the best of our knowledge, this is the first study to compare 3D hip kinematics and dynamics during straight walking and 90° turns while walking in healthy individuals and those with mild-to-moderate HOA.

We expected turning to require larger frontal and transverse hip mobility and provoke larger hip joint moments in these movement planes. In line with this hypothesis, both 90° turns required larger hip external and internal rotation and overall transverse hip RoM. In contrast to our expectations, frontal hip RoM was lower (step turn) or not different (spin turn) than during straight walking. However, there was reduced adduction but increased abduction during the step turn, and reduced abduction but increased adduction during the spin turn (Figure 6.2). As hypothesised, peak external hip joint moments in all movement planes were higher during both 90° turns than during straight walking, except for the peak hip adduction moment during the step turn. Thereby, observed joint moments are comparable to those reported for running (Schache et al., 2011), even though both turns are conducted with a lower CoM velocity than straight walking.

Secondly, we expected mobility deficits and modifications of joint loading of individuals with HOA to be more pronounced during turning than during straight walking. Individuals with HOA demonstrated decreased transverse RoM across all movements compared to healthy controls. Thereby, group differences in transverse RoM and peak hip internal rotation are largest during the 90° step turn (Figure 6.4). While Taylor et al. (2005) also reported larger external rotation angles during turning than straight walking, they did not find increased hip internal rotation during turning. These differences in hip rotation might stem from differences in foot placement strategies used by younger and older individuals. This warrants further investigation.

In the frontal plane, HOA-induced limitations in hip adduction and abduction were most noticeable during the spin and step turn, respectively (Figure 6.4). Previous studies found no differences in peak hip adduction or abduction during straight walking between healthy individuals and those with mild or moderate HOA (Constantinou et al., 2017; Rutherford et al., 2015a), and only partly for those with severe HOA (Hurwitz et al., 1997; Rutherford et al., 2015a; Tateuchi et al., 2014; Zügner et al., 2018). Likewise, no overall effect of HOA on frontal plane kinematics was found in this study. Thereby, frontal hip RoM during straight walking

as well as gait velocity observed in the present study were comparable to those reported by Rutherford et al. (2015a), with about 9° and 1.33 m/s. However, significant interaction effects in our study showed that group differences in peak hip adduction and abduction were most noticeable during spin and step turns, respectively. Tateuchi et al. (2014) also reported reduced abduction during a 45° step turn and reduced adduction during a 45° spin turn. However, participants in their study suffered from end-stage HOA. Thus, analysing 90° step and spin turns might allow detection of frontal plane hip mobility limitations at an earlier disease stage, and thus facilitate earlier treatment initiation.

Although joint moments were larger during the 90° turning movements, no overall group differences or interactions were found. Yet, large individual heterogeneity was observed for the joint moments during both 90° turns (Figure 6.2). These differences are likely to stem from different movement strategies adopted to perform the change in direction as rapidly as possible. Turning usually requires a leaning in motion to counteract the centripetal force (Courtine & Schieppati, 2003) which results in an increased distance between the base of support and the CoM. Previous studies found that older adults reduce leaning in motion in comparison to young adults (Yamaguchi et al., 2018) and that higher levels of knee extension strength are related to better maintenance of top-down segment reorientation and leaning-in behaviour in healthy elderlies (Dixon et al., 2018). Thus, individual differences in hip pain or strength ability likely influence trunk lean and subsequently hip joint moments. Further analyses of whole-body movement strategies could identify coping strategies adopted by individuals with HOA, and identify those with a more favourable hip joint loading to serve as a basis for gait retraining protocols.

For the sagittal plane, both turning movements required lower peak angles and less hip RoM than straight walking (Figure 6.2). Nevertheless, reduced peak hip extension leading to decreased sagittal RoM was observed across all movement tasks for individuals with HOA. Yet, group differences for sagittal hip RoM were largest during straight walking (Figure 6.4). Thus, limitations in sagittal plane hip mobility are best evaluated during straight walking.

Limitations

Results from this study have to be interpreted with regard to the following limitations. While transverse plane biomechanics represent important data for the analysis of turning movements, it has to be borne in mind that multi-body modelling is prone to errors in this plane. However, even if the modelling output might contain inaccuracies, observed differences between movement tasks and groups are large.

Secondly, observed hip angles in the sagittal plane are shifted towards lower flexion and more pronounced extension compared to the literature. This offset is caused by differences in the definition of the pelvis neutral position (Falisse et al., 2018; Roelker et al., 2017). Consequently, when comparing our data to other studies, individual model definitions should be considered.

Moreover, in our study, participants were asked to perform the turns in an abrupt manner, as previously done in other studies (Yamaguchi et al., 2018), to increase standardisation of the turning movement across trials and participants. However, performing the entire 90° turn within one GC might not reflect natural turning behavior (Akram et al., 2010; Bergsma et al., 2021). As joint moments (Xu et al., 2006), step width (Bergsma et al., 2021), and likely also joint angles increase with increasing turning angle, the peak angles and moments observed in our study might be larger than during turns in which the rotation of the body is split into multiple steps.

Further, to increase uniformity across multiple trials of the same subject, movement velocity was controlled using light barriers. While this approach allows the detection of changes in movement velocity immediately during data acquisition, it does not guarantee a consistent movement across all trials, as only the mean velocity between the light barriers is evaluated. Therefore, averaging peak values across different trials might not represent peak values observed during individual trials. Thus, our results likely represent hip mobility and joint loadings across different movement strategies, and future analyses of whole-body movements on single-trial data are needed to clarify the role of different movement strategies on mobility requirements and joint loading.

Likewise, our analysis of the implications of mild-to-moderate HOA was done on group level to detect overall changes in turning biomechanics in this population. However, individual differences in HOA severity or sex might influence movement execution, as has been previously shown for straight walking (Foucher, 2017; Steingrebe et al., 2023).

Lastly, in our study, joint kinematics were only analysed during ST of walking and turning to match the gait phase evaluated for joint dynamics. While this approach follows previous studies (Tateuchi et al., 2014; M. Taylor et al., 2005), analyses of joint kinematics during the entire GC may provide additional insights into turning movements.

6.6 Conclusion

In this study, we demonstrated that 90° turns require overall larger transverse hip mobility and task-specific increases in frontal hip mobility. Thereby, frontal and transverse plane mobility limitations of patients with mild-to-moderate HOA were more prominent during 90° turns than during straight walking. Thus, routine inclusion of 90° turns during gait analysis might allow the identification of frontal and transverse hip mobility deficits at an early disease stage of HOA and, therefore, earlier access to conservative treatment options. Lastly, both turns cause larger joint moments at the hip joint than straight walking, with large inter-individual variability. Identification of individual movement strategies could be used as a foundation for gait retraining interventions.

7 **Manuscript IV - Effect of hip osteoarthritis on postural control and stair walking biomechanics**

Slightly modified version of the paper published as:

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7.1 Abstract

Stair walking is more demanding than level walking, and differences have been found between steady-state stair walking and gait-to-stair or stair-to-gait transition. However, there is limited knowledge on steady-state stair walking biomechanics in people with HOA, as all existing data were obtained during gait-to-stair transition. The aim of the present study was to investigate the effects of mild-to-moderate HOA on hip and trunk biomechanics and dynamic balance during steady-state stair walking.

Hip and trunk biomechanics and dynamic balance during steady-state stair walking were assessed in 21 participants with mild-to-moderate HOA and 21 age-matched healthy controls using an optoelectronic motion capture system and a custom-built instrumented six-step staircase. Dynamic balance was assessed using the inverted pendulum model with an extrapolated CoM. Differences between the two groups were analysed with an independent t-test either on discrete parameters or on entire time curves using SPM.

The HOA group showed longer GC and ST durations during stair ascent and descent, and lower peak vertical CoM acceleration during stair descent vs. the control group. Trunk kinematics did not differ between groups. Lower frontal plane hip RoM and hip internal rotation moment, but greater anterior margin of stability (MoS) were observed during stair ascent in the HOA vs. control group.

Mild-to-moderate HOA reduced frontal hip mobility but did not affect mediolateral dynamic balance during steady-state stair walking. Despite a lack of alterations in sagittal hip or trunk kinematics or dynamics, greater anterior MoS in HOA vs. control participants showed a more stable but potentially less efficient motion during stair ascent. Lastly, lower vertical acceleration during stair descent might reduce impact and joint loading at IC. Future studies should analyse gait efficiency and joint loading, and the effect of HOA severity and muscle force capacity on steady-state stair walking biomechanics in people with HOA.

7.2 Introduction

Hip osteoarthritis is a progressive disease that is very common, especially in the elderly (Dagenais et al., 2009). Of the people aged 45 years or above, 28 % show radiographic signs of HOA and 9.7 % develop symptomatic HOA (Jordan et al., 2009). Hip osteoarthritis affects all joint structures (Block & Shakoor, 2009), and causes pain and loss of function (Salaffi et al., 2005). These functional limitations can significantly alter movement patterns and affect mobility. Therefore, a great portion of research on HOA-induced modifications to movement biomechanics has focused on level walking (Steingrebe et al., 2023). However, other everyday tasks, such as stair walking, impose higher demands than level walking, in terms of greater lower limb sagittal and frontal joint angles, hip and knee joint moments (Andriacchi et al., 1980; Nadeau et al., 2003; Riener et al., 2002), and hip joint contact forces (Bergmann et al., 2001). Previous studies have found lower movement velocity and higher ST duration during stair ascent and descent, and lower cadence and higher double-support time during stair descent in people with moderate-to-severe HOA compared to healthy controls (Hall et al., 2017; Meyer et al., 2016). During stair ascent, people with varying degrees of HOA showed lower peak hip angles and RoM in all directions and planes compared to healthy controls (Hall et al., 2017; Meyer et al., 2016; Popovic et al., 2021), except for greater peak hip external rotation in people with mild-to-moderate HOA compared to healthy controls (Popovic et al., 2021). Peak hip flexion, abduction and external rotation moments were also found to be lower in people with moderate-to-severe HOA compared to healthy controls during stair ascent (Hall et al., 2017; Meyer et al., 2016). During stair descent, people with moderate-to-severe HOA demonstrated lower sagittal hip RoM, peak hip extension and external rotation moment, but greater ipsilateral trunk lean compared to healthy controls (Hall et al., 2017).

However, these studies used staircases with only one or two steps (Hall et al., 2017; Meyer et al., 2016) or analysed the biomechanics on the first or last step (Popovic et al., 2021). Lower body kinematics of transition steps from level walking to stair walking or vice versa

differ from those during steady-state level or stair walking. Thereby, larger hip and knee flexion was found in healthy people during steady-state stair walking steps compared to transition steps (Alcock et al., 2014; Andriacchi et al., 1980; Neuman & Fey, 2023). In contrast to transition steps, anteroposterior positioning of the foot is restricted by tread size in steady-state stair walking (Bosse et al., 2012). Therefore, the base of support (BoS) during this task is likely smaller than during the transition, imposing higher demands on dynamic balance and CoM control (Novak et al., 2016). In static movements, postural stability is assumed if the projection of the CoM falls within the boundaries of the BoS. Yet, this is not always true in movements like walking, where the CoM is outside the BoS, but postural stability is maintained (Winter, 1995). Consequently, Hof (2008; 2005) developed the concept of the extrapolated CoM (xCoM), which considers both the position and velocity of the CoM. This concept has been used frequently to quantify balance during dynamic movement tasks, including step and stair walking (Bosse et al., 2012; Koyama et al., 2015; Novak et al., 2016). Previous studies employed a variety of balance tests to assess the impact of HOA on balance and have yielded conflicting results: Balance impairments of people with HOA were only found partially when compared to healthy controls or the contralateral limb (Picorelli et al., 2018). Thereby, most studies used static tasks to assess balance, which is not transferable to dynamic balance tasks, as balance is task-specific (Ringhof & Stein, 2018).

Therefore, we aimed to investigate steady-state stair walking biomechanics and task-specific dynamic balance in people with mild-to-moderate HOA. We hypothesised that HOA participants would show (I) lower movement velocity; (II) lower peak hip flexion angle and anterior trunk lean, increasing anteroposterior dynamic stability; (III) lower hip adduction angle but greater hip abduction angle, hip external rotation angle and ipsilateral trunk lean, increasing mediolateral dynamic stability; and (IV) lower external hip joint moments in all movement planes

7.3 Materials and Methods

7.3.1 Participants

Data for this case-control study were collected in the biomechanics laboratory of the Institute of Sports and Sports Science (IfSS) of the KIT (Germany) between April 2019 and March 2021. Participant inclusion and exclusion criteria are listed in Table 6.1.

In total, 42 individuals participated in this study: 21 with symptomatic mild-to-moderate HOA and 21 without HOA. Table 7.1 provides detailed participant characteristics for both groups. The HOA and control groups did not differ regarding age, body mass, height, BMI or functional leg length.

The study procedure was approved by the ethical committee of the KIT. All participants gave written informed consent for study participation and publication of anonymised results before study participation.

Table 7.1: Participant demographic and clinical characteristics.

	HOA group (n = 21)	Control group (n = 21)	p (t-test/MWU)
Male/Female [n]	11/10	11/10	
Age [years]	64.0 (9.6)	63.1 (9.2)	0.769
Body mass [kg]	71.3 (11.9)	74.4 (12.7)	0.429
Height [cm]	171.2 (6.7)	171.1 (8.8)	0.981
Body Mass Index [kg/m ²]	24.2 (2.9)	25.2 (2.7)	0.257
Harris Hip Score	74.6 (11.8)	98.4 (2.3)	< 0.001
Hip Osteoarthritis Outcome Score	62.0 (16.4)	97.7 (5.1)	< 0.001
Tegner Activity Score	4.7 (0.8)	4.9 (1.2)	0.370
Involved/analysed side [n]	11 right, 10 left	11 right, 10 left	
Kellgren-Lawrence Score [n]	Grade 2 = 9 Grade 3 = 7 Grade 3-4 = 1 Grade 4 = 4		
Functional leg length [cm]	79.0 (4.0)	79.5 (6.2)	0.748

Note. Data presented as mean with SD in parentheses. P-values as revealed by independent sample t-tests / Mann-Whitney-U tests (MWU); significant results ($p < 0.05$) in bold. HOA = hip osteoarthritis. Functional leg length = crotch height during upright standing (Choi & Ashdown, 2011).

7.3.2 Study procedure and data acquisition

Participants performed five valid trials (no handrail use, no tripping) of stair ascent and descent at a self-selected speed in a step-over manner. Self-selected movement speed was chosen to capture the natural movement strategies of people with and without HOA. Stair

ascent was initiated a few steps before the staircase, whereas descent was initiated without approach steps from the top platform.

Hip and trunk biomechanics were assessed using a 16-camera infrared motion capturing system (200 Hz; Vicon Motion Systems, Oxford Metrics Group, Oxford, UK) and a custom-built instrumented staircase with six steps (Figure 7.1). These had a riser height of 17 cm and a tread of 28 cm (inclination angle 31.26°). Steps four and five were equipped with 3D strain gauges (1000 Hz; ME-Meßsysteme GmbH, Henningsdorf, Germany), while the remaining steps were instrumented with 1D strain gauges for segmentation purposes (1000 Hz; Wii Balance Board, Nintendo, Kyoto, Japan). Participants were equipped with a full-body marker set (42 retroreflective markers; Figure A.22; Table A.4) by the same researcher, and several anthropometric measures were taken for individual scaling of the biomechanical model (Table A.5).



Figure 7.1: Picture of the instrumented staircase. Three-dimensional force plates in steps four and five from the bottom, one-dimensional force plates in steps one to three and six.

Note. Riser height = 17 cm, tread = 28 cm, inclination angle = 31.26° .

7.3.3 Data processing and analysis

Postprocessing of data was conducted in Vicon Nexus (version 2.14.0; Vicon Motion Systems, Oxford Metrics Group, Oxford, UK) and MATLAB (version R2022a; The MathWorks Inc., Natick, MA, USA). Kinematic and GRF data were filtered using a fourth-order Butterworth low-

pass filter with a cut-off frequency of 10 Hz (Kristianslund et al., 2012). 3D joint angles and external joint moments of the hip joint, and the CoM, were calculated using an inverse kinematics and dynamics approach with the multi-body model ALASKA Dynamicus (Hermsdorf et al., 2019), including the HJC definition proposed by Harrington et al. (2007). Hip joint moments were normalised to body mass. CoM calculation was based on a mathematical model including 20 segments (Figure A.23) of homogenous density, scaled using 65 individual anthropometrical measures (Table A.5). Total CoM was calculated using weighted summation using the previously calculated mass and inertia properties of the segments (Hermsdorf et al., 2019). Sagittal and frontal trunk angle was calculated as the angle between a vector from the midpoint of left and right ASIS markers to the clavicle marker and the vertical axis. The evaluated GC of the ipsilateral limb for the HOA group or a matched limb for the control group started with IC on the fourth step and ended with IC on the sixth step for stair ascent, or started with IC on the fifth step and ended with IC on the third step for descent. IC was detected by a vertical GRF > 15 N. The dependent variables were peak 3D hip joint angles and moments, mean vertical and anteroposterior CoM velocity, peak vertical and anteroposterior CoM acceleration, peak sagittal and frontal trunk angles, GC and ST duration.

Dynamic balance was assessed using the inverted pendulum model described by Hof et al. (2005) in the anterior, posterior, medial and lateral directions during the single support phase of the GC. Instability in this model is defined as the xCoM being outside the boundaries of the BoS. Thereby, xCoM is defined as the position of the CoM (pCoM) plus its velocity divided by the eigenfrequency of the pendulum, with the pendulum length (l) being defined as the instantaneous distance from the ipsilateral ankle joint centre to the CoM and with g being the acceleration of gravity.

$$xCoM = pCoM + vCoM / \sqrt{g/l}$$

Stability of the system is then expressed by the MoS, which is defined as:

$$MoS = BoS - xCoM.$$

Negative MoS values indicate an instantaneous instability of the system, which has to be counteracted, for example, by a change in BoS to prevent falling. Dynamic balance was assessed in the sagittal and frontal planes. The outer edge of the BoS was defined laterally by the fifth metatarsal marker, medially by the first metatarsal marker, anteriorly by the toe marker and posteriorly by the heel marker. If the foot was not positioned entirely on the tread, the edge of the BoS was defined by the tread edge (Novak et al., 2016). Minimum and

maximum MoS during the single-support phase in each direction were used as dependent variables.

All parameters were averaged across five valid trials of each movement. Angle, moment and MoS time curves were time-normalised to 100 % of GC, ST and single-support phase, respectively.

7.3.4 Statistical analysis

Statistical analyses of discrete parameters were performed in R (version 4.2.2; (R Core Team, 2023)) and the `rstatix`, `effsize`, `psych` and `stats` packages. Normal distribution and homogeneity of variance were assessed using Shapiro-Wilk and Levene's tests, respectively. Differences between groups were assessed using independent sample t-tests or Mann-Whitney U tests (MWU). In a secondary analysis, group comparisons were verified using an ANCOVA with leg length as a covariate. Since no discrepancies were found, t-test results are presented in the results section. The level of significance was set *a priori* to $p < 0.05$. Cohen's *d* (Cohen, 1988) was used to calculate effect sizes for all group comparisons. Additionally, continuous angle, moment and MoS time curves were compared using SPM or statistical nonparametric mapping (SnPM) (Pataky, 2010). To calculate SPM on observed time series, one trial of each participant was selected based on the lowest root mean square error to the average time curve across all five trials. SPM analyses were implemented using the open-source `spm1d` code (version M.0.4.10; `spm1d.org`) in MATLAB (version R2022a; The MathWorks Inc., Natick, MA, USA).

7.4 Results

Table 7.2 and Table 7.3 show descriptive data for all analysed parameters for stair ascent and descent, respectively. Figure 7.2 to Figure 7.5 present results of time curve analyses for hip joint angles and moments. The Appendix (Figure A.24 to Figure A.27) contains results of time curve analyses for trunk angles and MoS.

The HOA group ascended stairs with a longer ST ($p = 0.040$) and GC duration ($p = 0.040$) than the control group.

During stair descent, the HOA group had a longer ST ($p = 0.047$) and GC duration ($p = 0.039$), and lower peak vertical CoM acceleration ($p = 0.046$) compared to the control group.

7.4.1 Effect of hip osteoarthritis on sagittal hip and trunk motion and anteroposterior dynamic balance

During stair ascent, neither peak hip flexion angle ($p = 0.614$) nor sagittal trunk motion differed between groups (backward lean: $p = 0.690$; forward lean: $p = 0.746$). However, there was greater maximal anterior MoS in the HOA group ($p = 0.018$) than in the control group. SPM analysis showed a supra-threshold cluster for anterior MoS during stair ascent from 0 to 96 % single-support phase, with higher values for the HOA group ($p = 0.025$; Figure A.26).

During stair descent, neither peak hip flexion angle ($p = 0.424$) nor sagittal trunk motion (backward lean: $p = 0.189$; forward lean: $p = 0.276$) differed between groups. Likewise, no difference in anteroposterior dynamic balance between the HOA and control group was found (min. MoS anterior: $p = 0.052$; min. MoS posterior: $p = 0.202$; max. MoS anterior: $p = 0.073$; max. MoS posterior: $p = 0.165$).

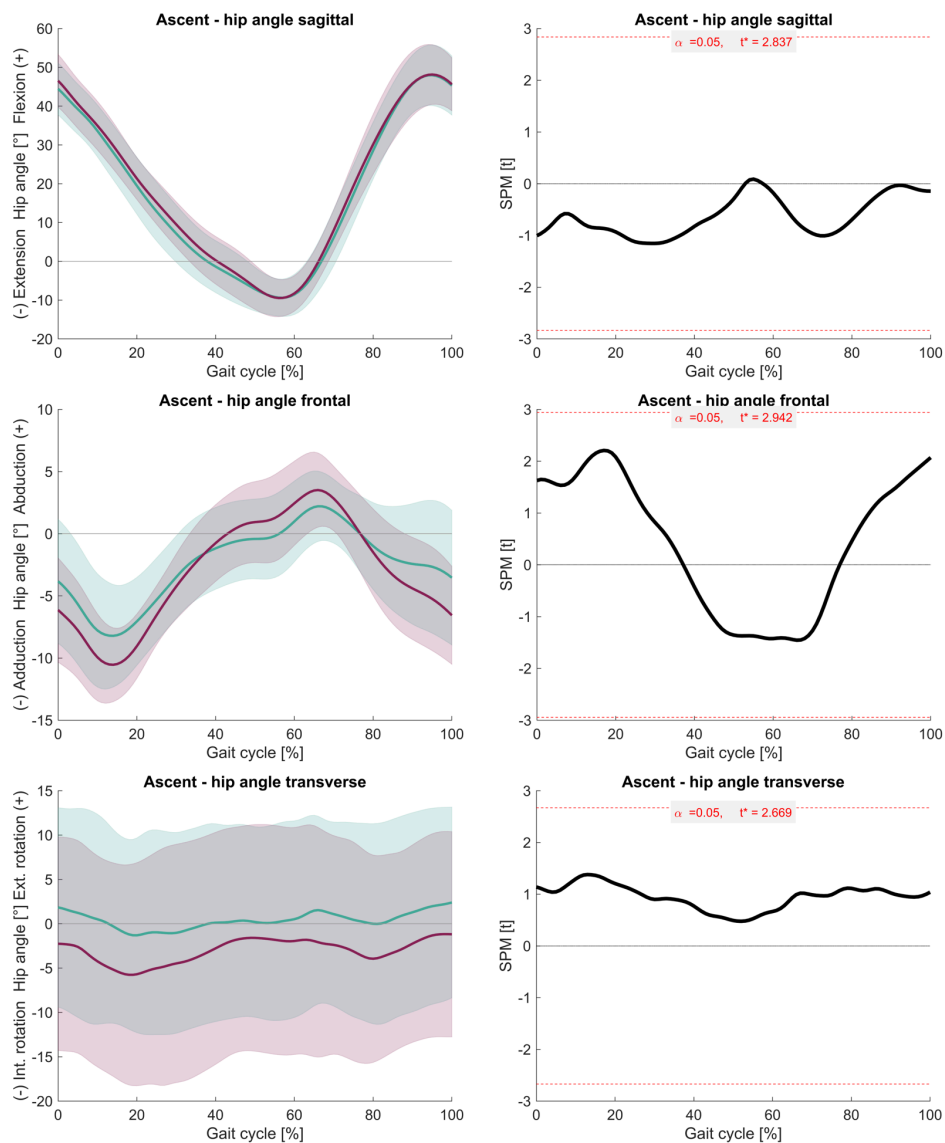


Figure 7.2: Hip angle time curves during stair ascent with results from SPM analysis.

Note. Data presented as mean $\pm$ SD. Green line = hip osteoarthritis group; purple line = control group.

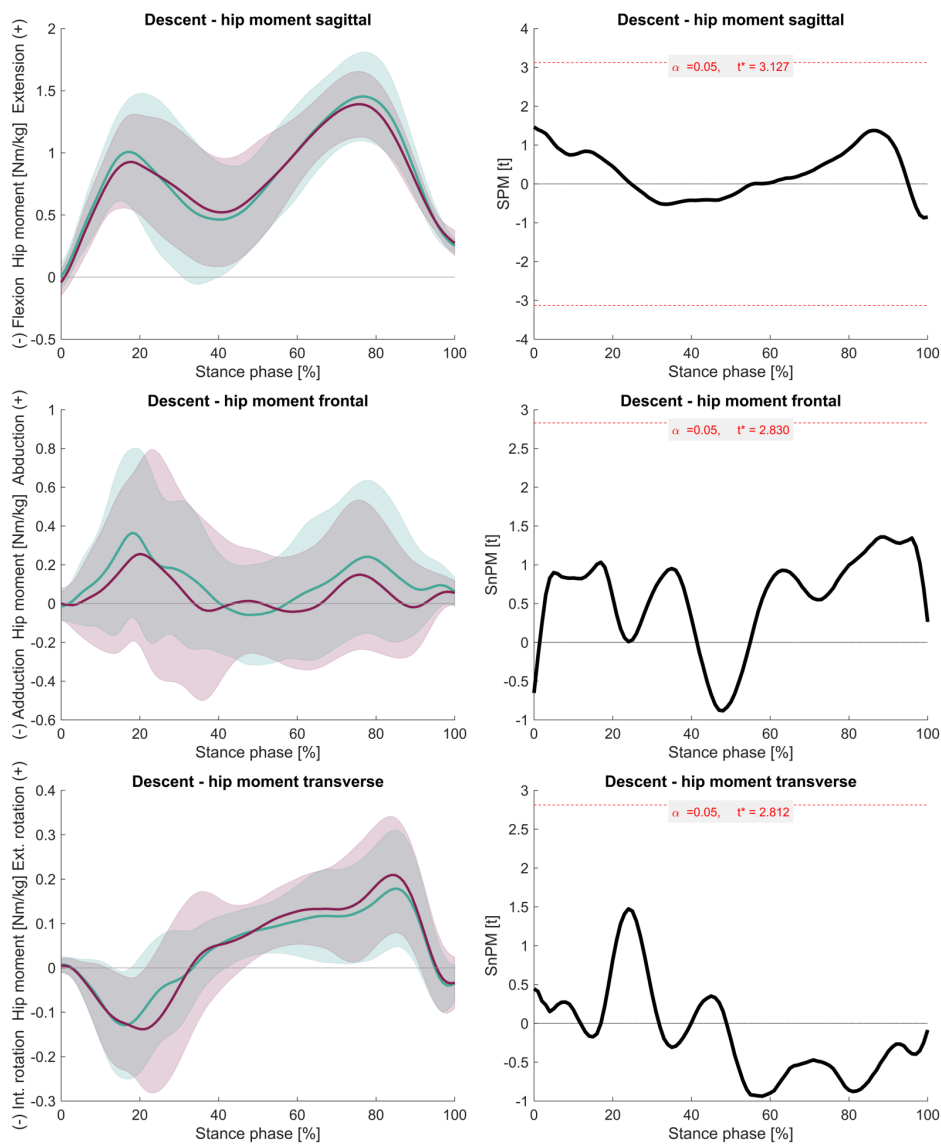


Figure 7.3: Hip angle time curves during stair descent with results from SPM and SnPM analyses.
Note. Data presented as mean $\pm$ SD. Green line = hip osteoarthritis group; purple line = control group.

Table 7.2: Biomechanical parameters during stair ascent.

Parameter during stair ascent		HOA group	Control group	Mean difference	t-test/MWU
		Mean (SD)	Mean (SD)	(95%CI)	p (d)
Sagittal plane	Hip peak extension [°]	-9.94 (4.87)	-9.78 (4.80)	0.16 (-2.86 3.18)	0.915 (0.03)
	Hip peak flexion [°]	47.76 (7.67)	48.92 (7.02)	1.15 (-3.43 5.74)	0.614 (0.16)
	Hip sagittal RoM [°]	57.70 (5.73)	58.69 (5.36)	0.99 (-2.47 4.45)	0.565 (0.18)
	Hip peak extension moment [Nm/kg]	0.54 (0.37)	0.38 (0.40)	-0.16 (-0.40 0.09)	0.087 (0.40)
	Hip peak flexion moment [Nm/kg]	-1.14 (0.40)	-1.32 (0.41)	-0.18 (-0.43 0.08)	0.167 (0.43)
	Peak backward trunk lean [°]	-3.34 (13.97)	-1.72 (16.46)	1.62 (-7.90 11.14)	0.690 (0.11)
	Peak forward trunk lean [°]	0.53 (14.19)	2.08 (16.36)	1.55 (-8.00 11.10)	0.746 (0.10)
	MoS minimum anterior [cm]	-15.60 (4.00)	-17.80 (3.00)	-2.20 (-4.37 0.04)	0.054 (0.61)
	MoS minimum posterior [cm]	14.90 (2.80)	16.10 (2.60)	1.20 (-0.47 2.94)	0.330 (0.45)
	MoS maximum anterior [cm]	4.00 (3.20)	1.60 (2.90)	-2.40 (-4.26 -0.43)	0.018 (0.76)
	MoS maximum posterior [cm]	34.40 (3.50)	35.50 (2.60)	1.10 (-0.81 3.02)	0.250 (0.36)
Frontal plane	Hip peak adduction [°]	-8.92 (3.96)	-10.96 (3.14)	-2.04 (-4.27 0.18)	0.071 (0.57)
	Hip peak abduction [°]	3.54 (2.85)	4.33 (2.50)	0.80 (-0.88 2.47)	0.342 (0.30)
	Hip frontal RoM [°]	12.45 (3.45)	15.29 (4.08)	2.84 (0.48 5.20)	0.019 (0.75)
	Hip peak adduction moment [Nm/kg]	-0.47 (0.22)	-0.50 (0.26)	-0.03 (-0.18 0.12)	0.670 (0.13)
	Hip peak abduction moment [Nm/kg]	0.20 (0.14)	0.28 (0.19)	0.08 (-0.02 0.19)	0.160 (0.50)
	Peak contralateral trunk lean [°]	-2.45 (1.23)	-2.23 (2.25)	0.22 (-0.91 1.35)	0.901 (0.12)
	Peak ipsilateral trunk lean [°]	2.40 (2.34)	2.25 (2.09)	-0.16 (-1.54 1.23)	0.862 (0.07)
Frontal plane	MoS minimum medial [cm]	-3.10 (2.10)	-3.10 (1.40)	0.00 (-1.05 1.20)	0.893 (0.04)
	MoS minimum lateral [cm]	8.30 (1.00)	8.30 (0.80)	0.00 (-0.54 0.63)	0.880 (0.05)

Table 7.2: Biomechanical parameters during stair ascent.

Parameter during stair ascent		HOA group	Control group	Mean difference	t-test/MWU
		Mean (SD)	Mean (SD)	(95%CI)	p (d)
	MoS maximum medial [cm]	1.90 (1.20)	1.80 (0.80)	-0.10 (-0.71 0.56)	0.806 (0.08)
	MoS maximum lateral [cm]	13.20 (2.00)	13.00 (1.30)	-0.20 (-1.28 0.84)	0.675 (0.13)
Transverse plane	Hip peak int. rotation [°]	-4.55 (10.79)	-7.89 (12.50)	-3.34 (-10.62 3.94)	0.360 (0.29)
	Hip peak ext. rotation [°]	5.79 (11.14)	2.42 (11.94)	-3.38 (-10.58 3.83)	0.349 (0.29)
	Hip transverse RoM [°]	10.35 (1.84)	10.31 (3.05)	-0.04 (-1.61 1.53)	0.636 (0.01)
	Hip peak internal rotation moment [Nm/kg]	-0.13 (0.08)	-0.19 (0.09)	-0.06 (-0.11 -0.01)	0.021 (0.69)
	Hip peak external rotation moment [Nm/kg]	0.17 (0.06)	0.15 (0.05)	-0.02 (-0.05 0.02)	0.335 (0.30)
Spatiotemporal	Stance phase duration [s]	0.80 (0.14)	0.72 (0.07)	-0.07 (-0.14 0.00)	0.040 (0.66)
	Gait cycle duration [s]	1.24 (0.19)	1.14 (0.11)	-0.10 (-0.20 -0.01)	0.040 (0.66)
	Mean anteroposterior CoM velocity [m/s]	0.47 (0.07)	0.50 (0.05)	0.03 (-0.01 0.07)	0.092 (0.53)
	Mean vertical CoM velocity [m/s]	0.30 (0.04)	0.31 (0.03)	0.02 (-0.01 0.04)	0.157 (0.45)
	Peak anteroposterior CoM acceleration [cm/s ²]	0.50 (0.20)	0.40 (0.10)	-0.10 (-0.13 0.06)	1.000 (0.21)
	Peak vertical CoM acceleration [cm/s ²]	2.50 (0.70)	2.30 (0.50)	-0.20 (-0.59 0.17)	0.271 (0.34)

Note. 95%CI = 95 % confidence intervals; MoS = margin of stability; HOA = hip osteoarthritis; CoM = centre of mass. P-values as found by independent sample t-tests/Mann-Whitney-U tests (MWU); significant results ($p < 0.05$) in bold.

7.4.2 Effect of hip osteoarthritis on frontal hip and trunk motion and mediolateral dynamic balance

During stair ascent, peak hip adduction ($p = 0.071$) and abduction ($p = 0.342$) angles did not differ between groups. However, lower frontal hip RoM was found in the HOA group compared to the control group ($p = 0.019$). Peak hip external rotation angle ($p = 0.349$), ipsilateral trunk lean ($p = 0.862$) and mediolateral MoS parameters (min. MoS medial: $p = 0.893$; min. MoS lateral: $p = 0.880$; max. MoS medial: $p = 0.806$; max. MoS lateral: $p = 0.675$) did not differ between groups.

During stair descent, peak hip adduction ($p = 0.196$) and abduction ($p = 0.421$) angles did not differ between groups. Likewise, peak hip external rotation angle ($p = 0.504$), ipsilateral trunk lean ($p = 0.615$) and mediolateral MoS parameters (min. MoS medial: $p = 0.606$; min. MoS lateral: $p = 0.211$; max. MoS medial: $p = 0.163$; max. MoS lateral: $p = 0.271$) did not differ between groups.

7.4.3 Effect of hip osteoarthritis on external hip joint moments

During stair ascent, the HOA group had a lower peak internal rotation moment compared to the control group ($p = 0.021$).

During stair descent, no differences in hip joint moments were found between the HOA and the control group.

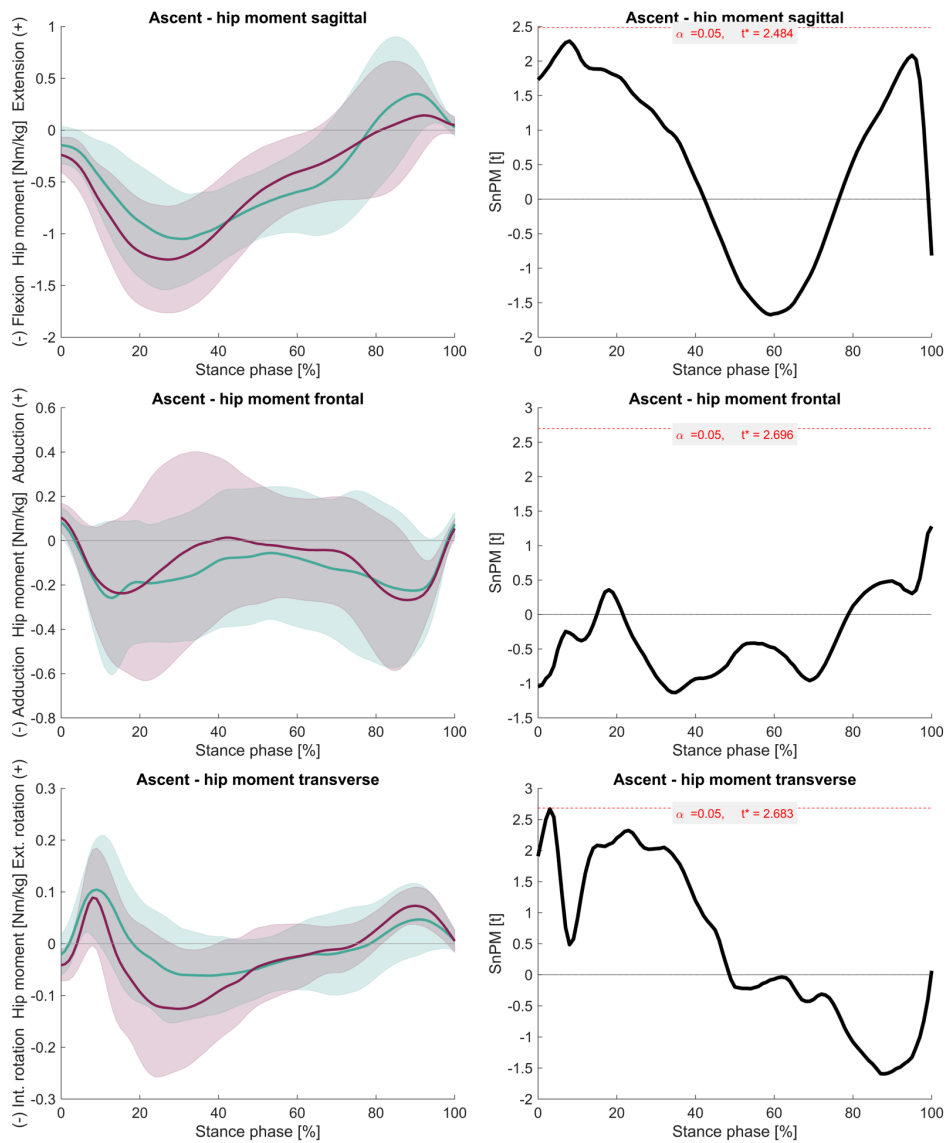


Figure 7.4: Hip moment time curves during stair ascent with results from SnPM analysis.
Note. Data presented as mean $\pm$ SD. Green line = hip osteoarthritis group; purple line = control group.

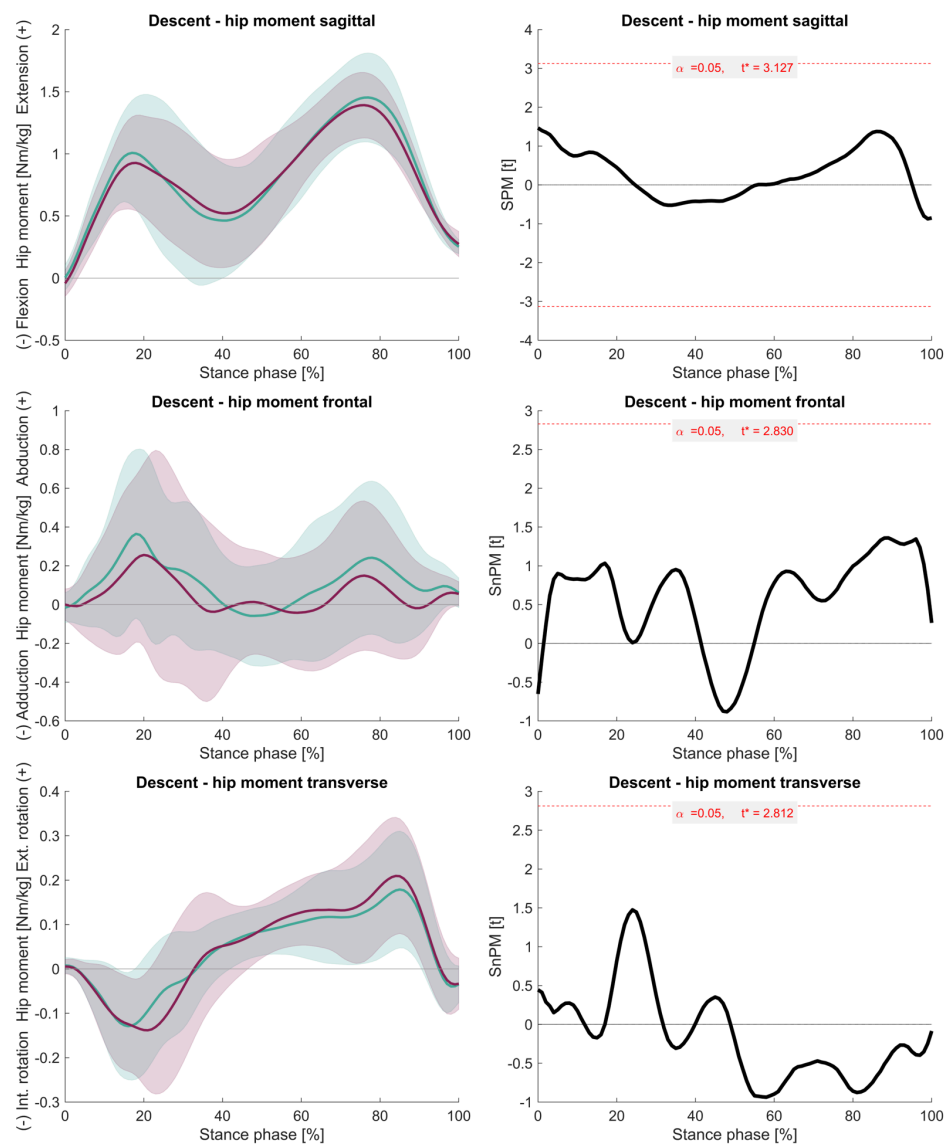


Figure 7.5: Hip moment time curves during stair descent with results from SPM and SnPM analyses.
 Note. Data presented as mean $\pm$ SD. Green line = hip osteoarthritis group; purple line = control group.

Table 7.3: Biomechanical parameters during stair descent.

Parameter during stair descent		HOA group	Control group	Mean difference	t-test/MWU
		Mean (SD)	Mean (SD)	(95%CI)	p (d)
Sagittal plane	Hip peak extension [°]	-6.67 (6.12)	-5.87 (4.85)	0.81 (-2.64 4.25)	0.638 (0.15)
	Hip peak flexion [°]	23.01 (9.85)	25.20 (7.60)	2.19 (-3.29 7.68)	0.424 (0.25)
	Hip sagittal RoM [°]	29.68 (5.84)	31.06 (4.17)	1.38 (-1.78 4.55)	0.382 (0.27)
	Hip peak extension moment [Nm/kg]	1.51 (0.34)	1.45 (0.21)	-0.06 (-0.23 0.12)	0.516 (0.20)
	Hip peak flexion moment [Nm/kg]	-0.14 (0.18)	-0.12 (0.13)	0.03 (-0.07 0.12)	0.960 (0.16)
	Peak backward trunk lean [°]	0.47 (3.77)	2.27 (4.87)	1.80 (-0.92 4.51)	0.189 (0.41)
	Peak forward trunk lean [°]	5.24 (3.79)	6.81 (5.31)	1.57 (-1.30 4.45)	0.276 (0.34)
	MoS minimum anterior [cm]	-20.00 (2.70)	-21.70 (3.00)	-1.70 (-3.54 0.02)	0.052 (0.62)
	MoS minimum posterior [cm]	24.30 (4.10)	26.10 (3.80)	1.80 (-0.65 4.33)	0.202 (0.46)
	MoS maximum anterior [cm]	-1.30 (3.60)	-3.20 (2.80)	-1.90 (-3.87 0.18)	0.073 (0.57)
	MoS maximum posterior [cm]	42.20 (3.00)	43.70 (3.70)	1.50 (-0.63 3.59)	0.165 (0.44)
Frontal plane	Hip peak adduction [°]	-6.80 (2.82)	-8.10 (3.58)	-1.31 (-3.31 0.70)	0.196 (0.41)
	Hip peak abduction [°]	4.42 (2.76)	5.08 (2.44)	0.65 (-0.97 2.28)	0.421 (0.25)
	Hip frontal RoM [°]	11.22 (4.03)	13.18 (3.52)	1.96 (-0.40 4.32)	0.132 (0.52)
	Hip peak adduction moment [Nm/kg]	-0.34 (0.23)	-0.42 (0.27)	-0.08 (-0.23 0.08)	0.230 (0.32)
	Hip peak abduction moment [Nm/kg]	0.57 (0.28)	0.60 (0.34)	0.03 (-0.16 0.22)	0.747 (0.10)

Table 7.3: Biomechanical parameters during stair descent.

Parameter during stair descent		HOA group	Control group	Mean difference	t-test/MWU
		Mean (SD)	Mean (SD)	(95%CI)	p (d)
Frontal plane	Peak contralateral trunk lean [°]	-1.61 (1.29)	-1.27 (2.24)	0.34 (-0.79 1.48)	0.550 (0.19)
	Peak ipsilateral trunk lean [°]	1.58 (1.60)	1.28 (2.24)	-0.30 (-1.52 0.91)	0.615 (0.16)
	MoS minimum medial [cm]	-5.60 (2.20)	-6.00 (2.40)	-0.40 (-1.80 1.06)	0.606 (0.16)
	MoS minimum lateral [cm]	9.20 (1.60)	9.70 (1.70)	0.50 (-0.53 1.55)	0.211 (0.31)
	MoS maximum medial [cm]	0.30 (1.00)	-0.10 (1.20)	-0.40 (-1.15 0.20)	0.163 (0.44)
	MoS maximum lateral [cm]	14.80 (2.10)	15.20 (2.90)	0.40 (-1.16 1.98)	0.271 (0.16)
Transverse plane	Hip peak int. rotation [°]	-3.96 (9.90)	-7.54 (14.02)	-3.58 (-11.15 3.98)	0.344 (0.30)
	Hip peak ext. rotation [°]	9.20 (11.10)	6.60 (13.70)	-2.60 (-10.37 5.18)	0.504 (0.21)
	Hip transverse RoM [°]	13.15 (4.58)	14.14 (4.76)	0.99 (-1.92 3.90)	0.496 (0.21)
	Hip peak internal rotation moment [Nm/kg]	-0.19 (0.11)	-0.22 (0.11)	-0.03 (-0.10 0.04)	0.330 (0.27)
	Hip peak external rotation moment [Nm/kg]	0.22 (0.11)	0.26 (0.10)	0.04 (-0.03 0.10)	0.256 (0.36)
Spatio-temporal	Stance phase duration [s]	0.70 (0.12)	0.63 (0.07)	-0.06 (-0.13 0.00)	0.047 (0.63)
	Gait cycle duration [s]	1.09 (0.19)	0.99 (0.11)	-0.10 (-0.20 -0.01)	0.039 (0.66)
	Mean anteroposterior CoM velocity [m/s]	0.53 (0.09)	0.57 (0.07)	0.04 (-0.01 0.09)	0.079 (0.56)
	Mean vertical CoM velocity [m/s]	0.32 (0.06)	0.35 (0.04)	0.03 (0.00 0.06)	0.068 (0.58)
	Peak anteroposterior CoM acceleration [cm/s ²]	0.70 (0.20)	0.70 (0.20)	0.00 (-0.08 0.12)	0.306 (0.11)
	Peak vertical CoM acceleration [cm/s ²]	1.40 (0.30)	1.60 (0.30)	0.20 (0.00 0.40)	0.046 (0.63)

Note. 95%CI = 95 % confidence intervals; MoS = margin of stability; HOA = hip osteoarthritis; CoM = centre of mass. P-values as found by independent sample t-tests/Mann-Whitney-U tests (MWU); significant results ($p < 0.05$) in bold.

7.5 Discussion

This was the first study to analyse biomechanics and task-specific dynamic balance during steady-state stair walking in participants with mild-to-moderate HOA. We found that during stair ascent, the HOA group had longer GC and ST duration, greater anterior MoS, but lower frontal hip RoM and hip external rotation moment compared to the control group. For stair descent, longer ST and GC duration, and lower peak vertical CoM acceleration were found in the HOA group compared to the control group.

7.5.1 People with hip osteoarthritis ascend and descend stairs with lower movement velocity

As hypothesised, movement velocity was lower in the HOA group than in the control group for both movement tasks, with longer ST and GC durations. This finding is in line with results from Hall et al. (2017), who also found longer ST duration during stair ascent and descent on a one-step staircase, and thus seems to be a consistent gait feature in people with HOA during gait-to-stair transition and steady-state stair walking. However, no reductions in antero-posterior and vertical CoM velocity were found between groups. While this aligns with the results from Popovic et al. (2021), it contradicts those of Hall et al. (2017). However, Hall et al. (2017) included participants with more severe HOA and calculated movement velocity by dividing stride length by stride time, as opposed to the CoM velocity used in our study. Popovic et al. (2021) did not report the method for velocity calculation, but values reported in their study were twice as high as in the present study or the study of Hall et al. (2017), and thus, are likely not comparable.

Additionally, we found lower peak vertical CoM acceleration during stair descent. This parameter has also been found to be lower in healthy older people than younger people (Buckley et al., 2013), caused by greater muscular co-contraction resulting in a stiffening of the limb. While greater muscular co-contraction reduces gait efficiency (Peterson & Martin, 2010), it might help to avoid high joint and muscle forces in the contralateral limb, which has to decelerate the CoM after IC. Analyses of muscle activity and joint contact forces during steady-state stair walking are needed to verify this hypothesis in people with HOA. If found valid, strengthening weakened hip and knee extensors might allow people with HOA to adopt a faster and more efficient movement pattern, especially during stair descent.

7.5.2 People with hip osteoarthritis ascend stairs with greater anteroposterior margin of stability

We expected to find differences in sagittal plane hip and trunk kinematics. In line with the results from Popovic et al. (2021) but contrary to Hall et al. (2017), we did not see an effect of HOA on sagittal hip RoM or peak hip flexion during stair ascent or descent. Hall et al. (2017) reported lower peak flexion and RoM during stair ascent. Overall, a greater sagittal hip RoM was described in their study, likely due to the larger riser height (20-24 cm vs. 17 cm in the present study), causing a greater hip flexion angle at IC. Yet, this contradicted previous literature's findings that reported larger hip flexion angles during steady-state stair walking compared to gait-to-stair transition (Alcock et al., 2014; Andriacchi et al., 1980). Lastly, participants in the study by Hall et al. (2017) had more severe HOA, which likely affected hip sagittal mobility (Baker et al., 2016).

Despite a lack of sagittal kinematic changes in our study, alterations in anterior stability were found. The HOA group had a greater peak anterior MoS and greater anterior MoS across nearly the entire ST of stair ascent in the SPM analysis compared to the control group. Therefore, participants with HOA seem to have adopted a more stable movement strategy during stair ascent. During steady-state stair walking, foot placement can only impact the anterior edge of the BoS to a small extent due to the dimensions of the tread, and in our study, no reduction in anteroposterior CoM velocity was found. Therefore, the greater MoS could either stem from a reduction in pendulum length, e.g. by greater knee flexion, or a combination of smaller modifications in CoM velocity, foot placement or pendulum length that did not reach statistical significance but still impacted the MoS. Yet, despite the greater MoS observed in the HOA group, both groups showed a negative anterior MoS throughout large parts of the single-support phase, which is necessary for an energetically efficient transition to the next step (Kuo et al., 2005). Therefore, the lower negative anterior MoS observed in the HOA group compared to the control group might have resulted from a movement strategy that reduced the risk of falling at the cost of reduced gait efficiency (Bosse et al., 2012).

7.5.3 Lower frontal hip mobility in people with hip osteoarthritis does not impact mediolateral dynamic balance

We expected the HOA group to show more pronounced hip abduction, hip external rotation and ipsilateral trunk lean to increase mediolateral postural stability during stair ascent and descent compared to the control group. In contrast to our hypothesis and previous literature findings (Hall et al., 2017; Popovic et al., 2021), no group differences in frontal plane peak angles during both movement tasks were found in the current study. However, the HOA group showed lower frontal hip RoM of about 20 % during stair ascent compared to the control group, similar to previous studies during gait-to-stair transition or level walking (Hall et al., 2017; Rutherford et al., 2015a). This reduction likely originated in lower hip adduction rather than abduction. As HOA causes decreased hip abduction strength (Loureiro et al., 2018) and GMed activity (Zacharias et al., 2020), reducing hip adduction is a strategy for lowering required hip abduction forces in people with HOA (Meyer et al., 2018). However, the lower frontal hip RoM did not translate into changes in mediolateral stability. Ylmaz & Bağcer (2022) found an effect of HOA severity on mediolateral but not anteroposterior stability, with more severe HOA causing worse stability scores during a stationary balance task. Therefore, mediolateral stability might only be impaired in people with more severe HOA and, consequently, larger impairments of hip abductor strength than observed in the participants of the present study. As the GMed not only serves as the main stabiliser of mediolateral CoM movement during stair ascent and descent, but also contributes substantially to the forward acceleration of the CoM during stair descent (Lin et al., 2015), HOA severity might also impact the anteroposterior CoM velocity, as discussed in chapter 7.5.1. Future studies must clarify the role of HOA severity and hip abductor strength on the mediolateral dynamic balance and the anteroposterior CoM velocity during stair walking.

7.5.4 People with hip osteoarthritis ascend stairs with lower hip internal rotation moments

We hypothesised that participants with HOA would show lower external hip joint moments in all movement planes. However, only lower peak hip internal rotation moments of about 32 % were found during stair ascent in the HOA group compared to the control group. This finding contradicts previous literature that found lower hip external rotation moments during stair ascent in HOA participants compared to control participants (Hall et al., 2017) or

found the reduction in hip external rotation moments to be the parameter with the highest explained variance during stair ascent in an HOA group (Meyer et al., 2016). However, both studies included individuals with more severe HOA than the present study. Peak internal rotation moments during stair ascent occur at around 30 % of the ST, where hip flexion is high. As the GMed also serves as a hip internal rotator, especially in hip flexion (Delp et al., 1999), lower abductor muscle force likely impacts transverse plane hip kinematics and dynamics. Yet, further investigation is warranted as force capacity and muscular activity were not captured during our study. For level walking, it has been shown that reductions in hip moments depend on HOA severity (Diamond et al., 2018). The results of our study imply that this could also be the case for stair walking.

7.5.5 Strengths and limitations

This was the first study to use a multi-step instrumented staircase to comprehensively assess steady-state stair walking biomechanics and dynamic balance in participants with mild-to-moderate HOA compared to a well-matched healthy control group. Thereby, not only discrete time points but also analyses of time series were conducted using SPM. Alongside these strengths of our study, some limitations have to be considered.

Angle and moment time curves often showed large SD, and thus, small group effects might have been hidden by large interindividual heterogeneity. While the sample size of the present study was comparable to or larger than in studies on people with more severe HOA (Hall et al., 2017; Meyer et al., 2016), it was smaller than in another study on people with mild-to-moderate HOA symptoms (Popovic et al., 2021). Therefore, larger sample sizes might be necessary to detect differences between groups in people with mild-to-moderate symptoms, where the disease impact on joint biomechanics is less pronounced than in people with more severe forms of HOA.

Participants in our study ascended and descended stairs at a self-selected speed, and our results showed lower movement velocity of the HOA group in terms of longer ST and GC duration, but not in terms of altered CoM velocity. As the gait velocity correlates with hip joint kinematics and dynamics (Fukuchi et al., 2019; Lelas et al., 2003), differences in movement speed might have impacted our results. However, self-selected movement velocity allows participants to keep their natural movement pattern and was also chosen in other studies (Hall et al., 2017; Popovic et al., 2021). We refrained from using an ANCOVA with movement velocity as a covariate, as differences in movement velocity are not a source

of random error variability but rather representative of the population characteristics (Astephen Wilson, 2012).

For the estimation of task-specific dynamic balance, an anatomical definition of the BoS based on markers was applied in the current study. Yet, this definition of the BoS can over- or underestimate the MoS (Curtze et al., 2024), as only about one-third of the physiological BoS can effectively be used for balance control during standing (Hof & Curtze, 2016). However, as the definition of the BoS was equal for both groups, this issue should not have influenced our findings or their interpretation.

Additionally, the interpretation of joint loading based on external joint moments is limited as muscle forces or forces of elastic structures are disregarded. Therefore, future studies should expand these findings using analysis of muscle activation and musculoskeletal modelling methods (Weinhandl & Bennett, 2019). Higher levels of muscular co-contraction have been reported in people with HOA (Diamond et al., 2020). They could also cause the reduced vertical CoM acceleration observed in our study, thus affecting joint loading.

7.6 Conclusion

We examined hip and trunk biomechanics and dynamic balance during steady-state stair walking in people with mild-to-moderate HOA. Despite the demanding nature of this task, no differences in trunk motion, and only a few alterations in hip biomechanics with lower frontal hip RoM and peak internal rotation moment during stair ascent were found in the HOA group compared to the control group. Yet, a greater anterior MoS showed that people with HOA modified their gait pattern during stair ascent to be more stable but potentially less efficient. Lastly, lower peak vertical CoM acceleration during stair descent was observed in the HOA group, which could have been caused by a stiffening strategy to reduce joint loading. Therefore, future studies must clarify the role of muscle strength and muscular co-contraction on steady-state stair walking biomechanics and gait efficiency in people with mild-to-moderate HOA.

8 Manuscript V - Effect of hip osteoarthritis and hip bracing on pain, functional capacity, proprioception and gait biomechanics

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8.1 Abstract

Hip Osteoarthritis is a common joint disease with a serious impact on the quality of life of the affected persons. Additionally, persons with HOA often show alterations in gait biomechanics. Developing effective conservative treatment strategies is of paramount importance, as joint replacement is only indicated for end-stage HOA. In contrast to KOA, little is known about the effectiveness of hip bracing for the management of HOA. Studies analysing mechanically unloading hip braces partly showed beneficial results. However, methodological limitations of these studies, such as small sample sizes or lack of control groups, limit the applicability of the results. Additionally, mechanically unloading braces might impose restrictions on motion and comfort and thus might not be suitable for people with only mild or moderate symptoms.

The aim of this study was to comprehensively quantify the effects of unilateral HOA as well as functional hip bracing on gait biomechanics, pain, proprioception and functional capacity in people with mild-to-moderate HOA.

Hip and pelvis biomechanics during walking were analysed in 21 individuals with mild-to-moderate HOA under three bracing conditions: unbraced, immediately after brace application and after one week of brace usage. Additionally, pain, hip proprioception and functional capacity were assessed. A matched group of 21 healthy individuals was included as reference. Kinematic and kinetic data were collected using a 16-camera infrared motion-capturing system and two force plates. Visual analogue scales, an angle reproduction test and a 6MWT were used to measure pain, hip proprioception and functional capacity, respectively.

Individuals with HOA walked slower, with reduced step length, sagittal hip RoM and peak extension angle and had a reduced functional capacity. After one week of brace application, step length, walking speed and functional capacity were significantly increased. Additionally, pain perception was significantly lower in the intervention period. These results encourage the application of functional hip braces in the management of mild-to-moderate HOA. However, as key parameters of HOA gait, such as a reduced peak extension angle, remained unchanged, the underlying mechanisms remain partly unclear and have to be considered in the future.

8.2 Introduction

Osteoarthritis is a degenerative joint disease with the hip joint being one of the most commonly affected joints (Litwic et al., 2013; Turkiewicz et al., 2014). The risk of developing HOA increases with age (Bijlsma & Knahr, 2007): about 28 % of people above the age of 45 show radiographic signs of HOA, and 9.7 % develop symptomatic HOA (Jordan et al., 2009). Due to demographic changes, the number of people suffering from HOA is likely to increase in the future (Fuchs et al., 2017). This stresses the need for effective treatment strategies as the emergence of HOA has serious consequences for the quality of life of the affected persons (Salaffi et al., 2005).

Previous studies have shown that people suffering from various degrees of HOA show altered gait kinematics, including a reduced walking speed, step length and cadence (Constantinou et al., 2017; Eitzen et al., 2012; Hulet et al., 2000; Hurwitz et al., 1997; Kubota et al., 2007; Ornetti et al., 2010; Watelain et al., 2001). Additionally, a reduced sagittal plane RoM as well as peak extension angle have been observed (Altman et al., 1991; Constantinou et al., 2017; Eitzen et al., 2012; Hurwitz et al., 1997; Kumar et al., 2015; Pua et al., 2009; Watelain et al., 2001) accompanied by a discontinuity of the hip extension movement (Foucher et al., 2012; Hurwitz et al., 1997). Frontal and transverse plane hip kinematics have been less frequently

analysed, but some studies reported reduced hip adduction angles (Leigh et al., 2016; Meyer et al., 2018; Tateuchi et al., 2014) and decreased hip internal rotation (Leigh et al., 2016). Furthermore, changes in pelvic kinematics have been found and interpreted as compensatory mechanisms for limited hip mobility (Murray et al., 1971; Watelain et al., 2001). In the sagittal plane, Murray et al. (1971), Kubota et al. (2007) and Leigh et al. (2016) showed an increased anterior tilt while Watelain et al. (2001) reported increased posterior tilt at push-off as well as an increased RoM. In the frontal plane, again, an increased RoM was found as well as an increased pelvis drop on the unsupported side (Watelain et al., 2001). Additionally, Murray et al. (1971), Thurston (1985) and Leigh et al. (2016) reported increased transverse pelvic rotation.

Hip joint loading studied by measuring joint moments revealed reduced external hip flexion and hip extension moments (Eitzen et al., 2012; Foucher, 2017; Hurwitz et al., 1997; Tateuchi et al., 2014; Watelain et al., 2001), reduced external hip adduction (Foucher, 2017; Hurwitz et al., 1997) and abduction moments (Kubota et al., 2007; Meyer et al., 2018) as well as reduced internal and external rotation moments (Hurwitz et al., 1997) for people with HOA. A recent meta-analysis highlighted that these reductions of joint moments cannot be solely attributed to a reduced walking speed but rather are the result of altered movement patterns (Diamond et al., 2018). Thereby, the reduction of joint moments seems to depend on HOA severity, with a reduction present in people with end-stage HOA but not in people with moderate HOA (Diamond et al., 2018).

In addition to gait impairments, several studies have shown impairments in postural control in people with HOA (Picorelli et al., 2018; Slomka et al., 2019; Truszczyńska et al., 2016), which have partly been attributed to a loss of proprioception (Slomka et al., 2019). Proprioception strongly relies on the afferent information of a group of mechanoreceptors (Gardner, 2021), of which there is a significantly lower number in the hip joint capsule, labrum and femoral head ligament of people with HOA (Moraes et al., 2011). Decreased joint position sense as one measure of proprioceptive acuity has been frequently reported for KOA patients (Barrett et al., 1991; Knoop et al., 2011) while, to our knowledge, no studies exist for HOA patients.

To date, it is still not clear whether these gait modifications solely reflect adaptations to pain and movement constraints or, to some extent, are provoking HOA and driving degeneration of the affected joint (Eitzen et al., 2012; Hurwitz et al., 1997; Leigh et al., 2016). However, persistent gait modifications might cause subsequent muscle weaknesses in less-used muscles (Hurwitz et al., 1997), causing a downward spiral of poor posture, pain and a decrease

in functional capacity. Therefore, treatment strategies should aim at pain relief and normalisation of the gait pattern to avoid secondary damage to adjoining soft tissue areas and joints (Hurwitz et al., 1997; van Drongelen, Wesseling, et al., 2020). Additionally, reduction of disease progression and increase of functional capacity should be strived to enhance the perceived quality of life (Perrot, 2012). Overall, conservative treatment strategies, such as exercise, are of paramount importance for the management of symptoms, as joint replacement is only indicated for end-stage OA (K. L. Bennell & Hinman, 2011; S. Murphy et al., 2016). However, in contrast to KOA, profound knowledge about conservative, non-pharmacological treatments for HOA is still sparse and current guidelines for HOA treatment often solely recommend exercise due to a lack of reliable data on other conservative treatment options (K. L. Bennell & Hinman, 2011). One aspect of conservative treatment strategies is the use of braces. While brace usage has been extensively studied (Beaudreuil et al., 2009; Cudejko et al., 2018; Duivenvoorden et al., 2015) and is often recommended for people with KOA (Kolasinski et al., 2020), only limited studies exist analysing their therapeutic effectiveness in people with HOA.

Most previously published studies analysed the effects of braces, aiming to mechanically unload the hip joint (Nérot & Nicholls, 2017; E. Sato et al., 2012; T. Sato et al., 2008; Shiba et al., 1998; Yamaji et al., 2009). Of these, the first (Shiba et al., 1998) studied the effects of the Hip Joint Moment Reduction Brace and showed that the brace was able to reduce hip abductor muscle activity during walking by 32.6 %. However, these results were observed only in a small population of healthy participants, and no inverse dynamic analysis of hip moments was conducted. Additionally, the authors stated that the use of such an unloading brace might impact the ability to conduct activities of daily living, which limits the area of application. Next, Sato and colleagues analysed the effects of the WISH brace, designed to restrict hip adduction and exert pressure on the greater trochanter. Brace wear resulted in pain relief and reduced the dependency on analgesics (E. Sato et al., 2012; T. Sato et al., 2008); however, due to a simultaneous walking exercise, the effects might not be solely attributable to brace application. Yamaji et al. (2009) analysed the effects of the WISH brace on GRF during gait and found an increase in the first peak of the vertical GRF. However, a very small sample size of seven participants was studied, and no joint kinematics or kinetics were reported, limiting the insights of these results. Another brace for individuals with HOA aims to reduce the internal hip abduction moment by externally applying hip abduction and external rotation forces (Nérot & Nicholls, 2017). Brace application resulted in a significant reduction of the peak adduction and internal rotation angles, as well as the peak internal abduction moment during the ST of level walking. Yet, the effects on pain perception varied strongly between

participants (Nérot & Nicholls, 2017). Additionally, only immediate brace effects have been studied, and data of the participants with HOA were not compared to a healthy control group to analyse normalisation of movement patterns due to brace application.

Although some of the analysed braces showed positive results on (e.g.) pain perception, hip braces are currently nearly exclusively used after hip arthroplasty to prevent excessive joint motion (Yonclas et al., 2006). Besides the generally sparse availability of studies, one reason for this might be the discomfort experienced by patients due to the weight and stiffness of hip braces designed to mechanically stabilise and unload the affected hip joint. Additionally, rigid braces that restrict motion might not be suitable for a population with moderate symptoms still engaging in sports or daily activities (Shiba et al., 1998).

Modern definitions of OA stress that it does not exclusively involve the joint cartilage but affects all joint structures (Block & Shakoore, 2009). Additionally, it has been shown for HOA that the correlation between symptoms and radiographic signs is inconsistent (Kinds et al., 2011). Therefore, brace concepts focusing on soft tissue joint structures, such as muscles or joint capsules, might be beneficial for the patients without the necessity of hard shells to exert force on the hip joint. As stated previously, current guidelines for HOA treatment often recommend the application of exercise, combined with physiotherapeutic treatments such as manual therapy and massage (Bannuru et al., 2019; Cibulka et al., 2017; Hernández-Molina et al., 2008; Kolasinski et al., 2020; The Royal Australian College of General Practitioners, 2018). Thus, applying conservative treatments such as friction and trigger point massage by a brace might, due to the longer application period, be beneficial for pain and joint stiffness, and subsequently allow HOA patients to enhance their physical activity level. Additionally, the use of elastic bandages and braces has been shown to increase the proprioceptive capacity in healthy individuals (Baltaci et al., 2011), in individuals with KOA (Barrett et al., 1991; Birmingham et al., 2001b), as well as in individuals without an anterior cruciate ligament (Birmingham et al., 2001a). However, results regarding the effectiveness for people with HOA are still lacking.

Therefore, the aims of the present study were first to analyse the effect of HOA on gait kinematics and dynamics, proprioception and functional capacity in a population with unilateral symptomatic HOA. Based on previous findings, we hypothesised that individuals with HOA would show a (I) reduced walking speed and step length; (II) reduced sagittal plane RoM and peak hip extension angle; (III) decreased hip proprioception; and (IV) decreased functional capacity compared to a healthy CON.

Secondly, we aimed to evaluate the short- and medium-term effects of hip brace application on gait kinematics and dynamics, pain, function and proprioception. We expected the brace application to (V) reduce pain perception, (VI) enhance hip proprioception, and (VII) increase functional capacity. For the hip kinematics, we expected (VIII) increased sagittal hip RoM and peak hip extension angle, resulting in (IX) longer step length and faster walking speed. While we expected (X) short-term brace application to have an immediate effect on hip proprioception, we expected (XI) the effects on functional capacity and gait biomechanics to arise only after medium-term brace application.

8.3 Materials and Methods

A case-control study and an intervention study were combined to investigate the research questions.

8.3.1 Participants

A total of 42 individuals participated in this study. The intervention group (HOA group) consisted of 21 individuals with unilateral symptomatic moderate HOA. The sample was an ad hoc convenience sample recruited through local physiotherapy practices and university information events concerning OA care. Assessment of HOA was based on clinical as well as radiographic criteria, as this provides higher sensitivity and specificity (Altman et al., 1991). Evaluation of the radiological images and classification of radiological OA signs based on the K-L-Score (Kellgren & Lawrence, 1957) was conducted by the same experienced orthopaedist.

The control group consisted of 21 healthy individuals without hip pain, matched to the mean age, weight, and height of the HOA group. Based on the involved side of the HOA group, eleven participants were randomly assigned to the right hip group, and ten participants to the left hip group.

The inclusion and exclusion criteria for the HOA and control groups are specified in Table 6.1. Details of the two groups are presented in Table 7.1

The study procedure was approved by the ethical committee of the KIT. All participants gave their written informed consent prior to study participation.

8.3.2 Data acquisition

To assess hip and pelvis biomechanics, three-dimensional gait analysis was conducted using a 16-camera infrared motion capturing system (Vicon, 200 Hz, Vicon Motion Systems, Oxford Metrics Group, Oxford, UK) and two 3D force plates (AMTI, 1000 Hz, BP600900, Advanced Mechanical Technology Inc., Watertown, MA, USA). A full-body marker set with 42 retro-reflective markers was used. Marker locations are specified in the Appendix (Figure A.22; Table A.4).

Sagittal plane hip proprioception was assessed using an active-active angle reproduction test. The testing procedure was based on the work of Arvin et al. (2015) with some structural changes to increase test feasibility and reliability (Steingrebe et al., 2019). Participants stood upright on a block of 20 cm height, allowing the ipsilateral leg to swing freely. Participants were blindfolded to eliminate any visual information and held onto a horizontal bar to facilitate balance. After a “Start” command, participants actively flexed their hip in a slow and steady manner until a “Stop” command was given by the instructor. When the participants felt to have sufficiently memorised the adopted position, they pressed a button, fixed at the handlebar, which powered an infrared light appearing in the simultaneous motion capture recordings and then returned to the initial position. After a rest period of three seconds, another “Start” command was given, and participants replicated the previously adopted position as precisely as possible, pushing the button when they felt they were in the correct position. A rest period of 30 seconds was given between each of the five trials.

As recommended by Cibulka et al. (2009), functional capacity was assessed using a 6MWT (Enright, 2003; Rejeski et al., 1995) on a standardised circuit of 54 m length completed in a counter-clockwise manner. Participants were asked to rate their current level of hip pain directly before and after completion of the test on a visual analogue scale of 10 cm length.

8.3.3 Study procedure

Participants of the HOA group were tested on three occasions. The time interval between sessions one and two was 5 ± 4 weeks, and between sessions two and three was 1 ± 0 week. Participants of the control group were tested on one occasion, which was identical to the procedure during the first session of the HOA group (Figure 8.1).

HOA group	Session 1	7 days before Session 2	Session 2	7 days before Session 3	Session 3
	Without brace	Baseline period	With brace short-term effects	Brace intervention period	With brace medium-term effects
	Inclusion/exclusion criteria; questionnaires (HHS, HOOS, TAS)	Daily pain assessment (pain during walking, night pain)	Brace fitting	Daily pain assessment (pain during walking, night pain)	
	Gait analysis		Gait analysis		Gait analysis
	Angle reproduction test		Angle reproduction test		Angle reproduction test
	6MWT		6MWT		6MWT

Control group	Session 1
	Without brace
	Inclusion/exclusion criteria; questionnaires (HHS, HOOS, TAS)
	Gait analysis
	Angle reproduction test
	6MWT

Figure 8.1: Study procedure for the hip osteoarthritis (HOA) and control group.
Note. HHS = Harris Hip Score; HOOS = Hip Osteoarthritis Outcome Score; TAS = Tegner Activity Score; 6MWT = 6-minute walking test.

Session 1: During the first session, participants were screened for the inclusion and exclusion criteria and completed two questionnaires concerning hip function (HHS (Harris, 1969) & HOOS (Klässbo, Larsson, et al., 2003)) and one questionnaire assessing their activity level (Tegner Activity Score (Swanenburg et al., 2014; Tegner & Lysholm, 1985)). Anthropometrical measures for biomechanical modelling, as well as fitting of the brace, were taken. Afterwards, participants were equipped with retroreflective markers. After a standardised instruction and a familiarisation trial, participants completed the proprioception test, followed by the biomechanical gait analysis. Thereby, participants walked barefoot at their self-selected speed for five valid trials (Laroche et al., 2011). For a valid trial, participants had to strike the first force plate entirely with the ipsilateral foot and the second force plate entirely with the contralateral foot. A constant walking speed was controlled using light barriers, allowing a difference of $\pm 5\%$ from the first valid trial. Finally, participants completed the 6MWT.

Baseline period: Participants recorded their pain perception during walking and at night using a standardised protocol with VAS scales on a daily basis for seven days.

Session 2: In the second session, participants were individually fitted with a hip brace (Cox-aTrain, Bauerfeind AG, Zeulenroda-Triebes, Germany) by an experienced orthopaedic technician. The hip brace consists of a pelvis belt equipped with a gluteal pad and two friction pads at the sacroiliac joints (SIJ), an aluminium joint splint including a moving trochanter ad and a thigh bandage (Figure 8.2). The brace aims at stabilising the pelvis through the tight-fitting pelvis belt and to stimulate several trigger points at the GMed and SIJ. Additionally, the moving trochanter pad applies a friction massage to the muscle insertions at the greater trochanter to relax hypertonic muscles and thereby increase hip joint mobility.



Figure 8.2: CoxaTrain hip brace.
Note. ©Bauerfeind AG

After a short period of brace familiarisation (autonomous walking with the brace throughout the lab until the participants felt comfortable), the procedure of session one was replicated. As the pelvis belt of the brace covers the anatomical landmarks for marker placement at the pelvis, retroreflective markers had to be placed on the brace (left and right ASIS) or were attached through the mesh fabric (left and right PSIS).

Brace intervention period: Participants were instructed to wear the brace during their daily activities for one week for at least four hours per day. Participants recorded the brace wear time on a standardised form and, additionally, the brace was equipped with a thermal sensor (*Orthotimer*[®], Rollerwerk Medical Engineering & Consulting, Balingen, Germany) to objectively detect brace wear time. Again, participants recorded their pain perception during walking and at night using a standardised protocol on a daily basis.

Session 3: The third test session was conducted in the same way as in session two.

8.3.4 Data processing and biomechanical modelling

Kinematic and GRF data of the walking trials were filtered using a 2nd-order Butterworth low-pass filter with a cut-off frequency of 15 Hz (Kristianslund et al., 2012). Kinematic data of the proprioception trials were filtered using a 2nd order Butterworth low-pass filter with a cut-off frequency of 6 Hz (Roithner et al., 2000).

An inverse kinematics and dynamics approach using the multi-body model ALASKA Dynamicus (Härtel & Hermsdorf, 2006) was used to calculate 3D joint angles and external joint moments. The location of the HJC was calculated based on the formula proposed by Harrington et al. (2007) and Kainz et al. (2015). Hip joint angles were defined as the rotation of the femur relative to the pelvis coordinate system (three Bryant angles). The axes of the pelvis coordinate system are defined by the two ASIS markers (1), the midpoints between the two ASIS markers and the two PSIS markers, and orthogonal to the plane through ASIS and PSIS markers (3).

Data post-processing was conducted using MATLAB (R2017b). Joint angles were time-normalised to one GC (100 time points) beginning with the heel strike on the force plate until the next consecutive heel strike of the same foot. Heel strike on the force plate was determined when the vertical GRF exceeded a threshold of 10 N (Tirosh & Sparrow, 2003). The consecutive heel strike was detected using the vertical trajectory of the heel marker. Joint moments were normalised to bodyweight as well as time normalised to the ST (100 time points). Normalised angle and moment curves were averaged across the five trials of each condition. Lastly, peak hip joint angles, hip joint RoM, as well as peak hip joint moments, were calculated and used as dependent variables.

Additionally, temporal-spatial gait parameters were analysed as dependent variables. Walking speed was calculated as the mean velocity of the CoM in the anterior-posterior direction across each GC. Stride and step length were calculated as the distance between the heel marker position at the first and second heel strike of the ipsilateral foot or at the first heel strike of the ipsilateral and contralateral foot, respectively, in the walking direction.

As a parameter for hip proprioception, the absolute angle error (AAE) was defined as the absolute difference between the hip flexion angles in the initially adopted and the resumed position (Arvin et al., 2015) (mean across 0.1 seconds from the instant of switch pressing).

Thus, lower error values represent a better hip proprioception. AAEs were calculated for each subject and averaged across the five trials.

Pain perception during walking as well as night pain were obtained daily via questionnaires using VAS in the baseline and intervention periods and averaged across the seven days.

8.3.5 Statistical analysis

All statistical tests were performed using IBM SPSS Statistics 25.0 (IBM Corporation, Armonk, NY, United States). All variables were tested for normal distribution using the Shapiro-Wilk-Test. If normal distribution could not be assumed, nonparametric statistical tests were applied.

Firstly, differences between the HOA group at baseline and control group (1 time point, 2 groups, 1 degree of freedom) were analysed using t-tests for independent samples (or MWU).

Secondly, comparisons within the HOA group under different bracing conditions (3 time points, 2 degrees of freedom) were conducted using univariate ANOVAs for repeated measures (or Friedman tests). If sphericity was violated, Greenhouse-Geisser estimates were used to correct for these violations. For significant results in the ANOVA, a post hoc analysis was conducted using t-tests for dependent samples (or Wilcoxon tests (WCX)) with Holm-Bonferroni corrections to adjust for multiple comparisons.

For all statistical tests, the level of significance was set *a priori* to 0.05. To estimate the effect sizes for between-group comparisons, Cohen's d was calculated based on means and SD. Effect sizes were interpreted as $|d| > 0.2$ being a small effect, $|d| > 0.5$ a moderate effect and $|d| > 0.8$ a large effect (Cohen, 1988). Effect sizes for comparisons within the HOA group were calculated using partial eta squared (small effect: $\eta_p^2 \geq 0.01$; medium effect: $\eta_p^2 \geq 0.06$; large effect: $\eta_p^2 \geq 0.14$) (Cohen, 1988; Richardson, 2011). For analyses conducted with the Friedman test, no effect sizes are presented.

Due to a technical error, the force plate only recorded GRF in the vertical direction for two participants in the control group. Therefore, these datasets were excluded during the analysis of joint moments. As the data can be classified as missing completely at random (Rubin, 1976) no corrections were made. Additionally, due to a technical issue, the thermal sensors

did not record the brace wear time for two participants. Calculation of average wear times was based on the questionnaire data for those two participants.

8.4 Results

There were no significant differences in the subject characteristics between the HOA and control groups concerning age, height, body mass, BMI or activity level (Table 7.1).

8.4.1 Effects of hip osteoarthritis on functional capacity, hip proprioception and gait biomechanics

The HOA group covered a significantly lower distance during the 6MWT (559.7 ± 84.4 m) compared to the control group (631.9 ± 54.8 m) ($p = 0.002$; $|d| = 1.01$). No significant group effects on the AAE during the angle reproduction test were found (HOA: $3.93 \pm 1.82^\circ$; CON: $3.80 \pm 2.00^\circ$; $p = 0.822$; $|d| = 0.07$).

Participants from the HOA group (1.23 ± 0.17 m/s) walked significantly more slowly than participants from the control group (1.32 ± 0.11 m/s) ($p = 0.049$; $|d| = 0.63$) and showed a significant reduction in step length (HOA: 0.62 ± 0.07 m; CON: 0.67 ± 0.06 m; $p = 0.013$; $|d| = 0.81$). There were no significant differences between groups regarding absolute (HOA: 0.64 ± 0.05 s; CON: 0.64 ± 0.05 s; $p = 0.774$; $|d| = 0.09$) and relative (HOA: 62.42 ± 1.93 %; CON: 62.9 ± 2.46 %; $p = 0.89$; $|d| = 0.22$) ST duration, absolute (HOA: 0.39 ± 0.02 s; CON: 0.38 ± 0.04 s; $p = 0.589$; $|d| = 0.28$) and relative (HOA: 37.58 ± 1.93 %; CON: 37.1 ± 2.46 %; $p = 0.89$; $|d| = 0.22$) swing phase duration or stride length (HOA: 1.26 ± 0.13 m; CON: 1.33 ± 0.15 m; $p = 0.061$; $|d| = 0.50$).

Results for kinematics and dynamics during walking are presented in Table 8.1. The HOA group showed a reduced RoM in the sagittal and transverse planes. Hip joint kinematics and moments were reduced in the sagittal plane (peak extension, peak hip extension and flexion moment) for the HOA group. Additionally, the HOA group showed a reduced peak adduction and peak external rotation moment. Finally, an increase in pelvic tilt RoM was observed in the HOA group.

In summary, as hypothesised, the HOA group showed (I) reduced walking speed and step length, (II) reduced sagittal hip RoM and peak extension angle, as well as (IV) reduced functional capacity. In contrast, no effects of HOA on (III) hip proprioception were found.

Table 8.1: Results of discrete hip & pelvis joint angles and hip joint moment gait parameters for the hip osteoarthritis (HOA) and control group (CON).

Parameter during level walking		HOA without brace	CON without brace	t-test/MWU
		Mean (SD)	Mean (SD)	p (d)
Sagittal plane	Hip peak extension [°]	-22.55 (6.28)	-29.10 (4.54)	< 0.001 (1.20)
	Hip peak flexion [°]	14.38 (5.84)	15.99 (5.54)	0.366 (0.28)
	Hip sagittal RoM [°]	36.94 (8.30)	45.09 (5.43)	0.001 (1.16)
	Peak ext. hip extension moment [Nm/kg]	0.53 (0.19)	0.65 (0.16)	0.038 (0.68)
	Peak ext. hip flexion moment [Nm/kg]	-0.69 (0.18)	-0.83 (0.17)	0.008 (0.80)
	Pelvic tilt RoM [°]	4.97 (2.22)	3.58 (0.87)	0.012 (0.82)
Frontal plane	Hip peak adduction [°]	-7.65 (2.30)	-9.04 (2.40)	0.063 (0.59)
	Hip peak abduction [°]	4.83 (2.47)	4.54 (2.47)	0.710 (0.12)
	Hip frontal RoM [°]	12.48 (2.97)	13.58 (2.20)	0.180 (0.42)
	Peak ext. abduction moment [Nm/kg]	0.19 (0.08)	0.20 (0.08)	0.592 (0.13)
	Peak ext. adduction moment [Nm/kg]	-0.92 (0.10)	-1.04 (0.15)	0.010 (0.95)
	Pelvic obliquity RoM [°]	6.31 (2.16)	6.57 (2.08)	0.687 (0.12)
Transverse plane	Hip peak internal rotation [°]	-6.71 (10.84)	-11.48 (13.06)	0.428 (0.40)
	Hip peak external rotation [°]	5.47 (10.34)	4.19 (13.07)	0.726 (0.11)
	Hip transverse RoM [°]	12.18 (3.08)	15.67 (3.28)	0.001 (1.09)
	Peak ext. internal rotation moment [Nm/kg]	-0.07 (0.04)	-0.10 (0.10)	0.379 (0.40)
	Peak ext. external rotation moment [Nm/kg]	0.20 (0.05)	0.25 (0.08)	0.035 (0.76)
	Pelvic rotation RoM [°]	10.41 (4.35)	10.79 (3.21)	0.359 (0.10)

Note. P-values as revealed by independent sample t-tests / Mann-Whitney-U tests (MWU) and effect sizes (Cohen's d); significant results ($p < 0.05$) in bold. RoM = range of motion.

8.4.2 Effects of hip bracing on pain, functional capacity, hip proprioception and gait biomechanics

8.4.2.1 Effects of hip bracing on pain, functional capacity and hip proprioception

During the intervention period, participants wore the brace on average for $10.1 (\pm 3.5)$ hours per day. The average perceived pain during walking activities significantly decreased during the intervention period (18.4 ± 18.1 mm) compared to the baseline period (25.7 ± 15.3 mm) ($p = 0.006$). Additionally, night pain was significantly reduced during the intervention period (13.9 ± 15.9 mm) compared to the baseline period (17.0 ± 17.6 mm) ($p = 0.042$). Pain reduction was reported by 18 out of 21 participants for walking activities and 14 out of 21 participants for night pain. For the participants with a positive brace effect on pain, mean reduction in VAS pain score was 10.5 ± 7.9 mm or 45.5 ± 28.3 % for walking activities and 6.2 ± 5.4 mm or 41.6 ± 30.7 % for night pain. For the participants with a negative effect on pain, the mean increase was 11.4 ± 8.2 mm or 40.8 ± 37.5 % for walking activities and 3.1 ± 3.2 mm or 84.2 ± 141.3 % for night pain.

The distance covered during the 6MWT significantly increased after medium-term brace application (589.1 ± 82.7 m) compared to the baseline condition (559.7 ± 84.4 m) ($p < 0.001$) as well as compared to short-term brace application (562.3 ± 80.9 m) ($p < 0.001$). The level of hip pain before ($p = 0.049$, post hoc analysis not significant) and after ($p = 0.363$) the 6MWT was not influenced by the bracing condition. No significant bracing effects on the AAE during the angle reproduction test were found (HOA without brace: 3.93 ± 1.82 ; HOA short-term: 3.37 ± 1.70 ; HOA medium-term: 3.57 ± 1.69 ; $p = 0.397$).

In summary, as hypothesised, brace application resulted in (V) reduced pain perception as well as (VII) increased functional capacity. As expected, the (XI) increase in functional capacity occurred only after medium-term brace application. In contrast, no brace effects on (V) pain during the 6MWT or on hip proprioception were found, neither (X) after short-term nor after medium-term brace application.

8.4.2.2 Effects of hip bracing on gait biomechanics

Effects of brace application on temporal-spatial gait parameters and discrete hip angle and hip moment parameters are displayed in Table 8.2 and Table 8.3, respectively. Joint angle and joint moment time curves are shown in Figure 8.3. Medium-term brace application resulted in a significant increase in gait velocity, stride and step length compared to the unbraced and short-term conditions. Additionally, ST duration decreased after medium-term brace application compared to the unbraced condition.

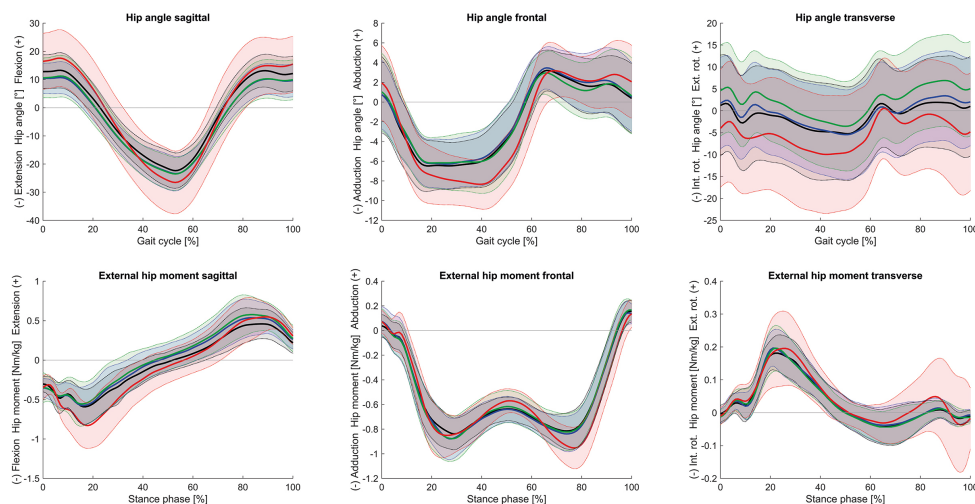


Figure 8.3: Time curves of hip joint angles [°] and moments [Nm/kg] normalised to gait cycle or stance phase.
Note. Data presented as mean $\pm$ SD. Red line = control group, black line = hip osteoarthritis (HOA) group without brace, blue line = HOA group short-term, green line = HOA group medium-term.

In the sagittal plane, brace application reduced the peak flexion angle after short-term application and increased the peak extension moment in both braced conditions. Additionally, pelvic tilt and pelvic rotation RoM increased with brace application.

In summary, as hypothesised, brace application resulted in (IX) an increase in step length and walking speed. As expected (XI), both changes occurred only after medium-term brace application. However, (VIII) no brace effects on hip sagittal RoM or peak extension angle were found, neither (XI) after short-term nor after medium-term brace application.

Table 8.2: Results of bracing effects on temporal-spatial gait parameters for the hip osteoarthritis group.

Parameter during level walking	Baseline	Short-term	Medium-term	ANOVA/ Friedman	B vs. S	B vs. M	S vs. M
	Mean (SD)	Mean (SD)	Mean (SD)	p (η_p^2)	p	p	p
Gait velocity [m/s]	1.23 (0.17)	1.26 (0.21)	1.31 (0.21)	0.001 (0.29)	0.153	0.003	0.032
Stance phase duration [s]	0.64 (0.05)	0.64 (0.05)	0.62 (0.06)	0.004†	0.244	0.006	0.06
Swing phase duration [s]	0.39 (0.02)	0.38 (0.02)	0.38 (0.02)	0.469 (0.03)			
Stance phase duration [%]	62.42 (1.93)	62.38 (1.78)	61.92 (1.88)	0.151 (0.10)			
Swing phase duration [%]	37.58 (1.93)	37.62 (1.78)	38.08 (1.88)	0.151 (0.10)			
Stride length [m]	1.26 (0.13)	1.27 (0.15)	1.30 (0.14)	0.003 (0.26)	0.271	0.003	0.032
Step length [m]	0.62 (0.07)	0.63 (0.08)	0.64 (0.07)	0.006 (0.23)	0.334	0.009	0.026

Note. P-values as revealed by one-way repeated measures ANOVAs / Friedman tests and Holm-Bonferroni corrected pairwise comparisons and effect sizes (η_p^2); significant results ($p < 0.05$) in bold; † marks analysis with a Friedman test. B = baseline; S = short-term; M = medium-term.

Table 8.3: Results of bracing effects on discrete kinematic and kinetic gait parameters for the hip osteoarthritis group.

Parameter during level walking	Baseline	Short-term	Medium-term	ANOVA/ Friedman	B vs. S	B vs. M	S vs. M
	Mean (SD)	Mean (SD)	Mean (SD)	p (η_p^2)	p	p	p
Hip peak extension[°]	-22.55 (6.28)	-23.69 (6.05)	-23.71 (5.80)	0.405†			
Hip peak flexion [°]	14.38 (5.84)	11.77 (5.47)	11.97 (7.14)	0.011 (0.20)	0.006	0.052	0.85
Hip sagittal RoM [°]	36.94 (8.30)	35.47 (8.34)	35.67 (8.76)	0.039 (0.17)	0.078	0.140	0.594
Hip peak extension moment [Nm/kg]	0.53 (0.19)	0.61 (0.21)	0.65 (0.21)	0.005 (0.26)	0.006	0.006	0.299
Hip peak flexion moment [Nm/kg]	-0.69 (0.18)	-0.64 (0.21)	-0.65 (0.25)	0.469 (0.04)			
Pelvic tilt RoM [°]	4.97 (2.22)	6.15 (2.53)	6.00 (2.47)	< 0.001 (0.32)	< 0.001	0.012	0.634
Hip peak adduction [°]	-7.65 (2.30)	-7.18 (2.38)	-7.37 (2.76)	0.432 (0.04)			

Table 8.3: Results of bracing effects on discrete kinematic and kinetic gait parameters for the hip osteoarthritis group.

Parameter during level walking	Baseline	Short-term	Medium-term	ANOVA/ Friedman	B vs. S	B vs. M	S vs. M
	Mean (SD)	Mean (SD)	Mean (SD)	p (η_p^2)	p	p	p
Hip peak abduction [°]	4.83 (2.47)	4.62 (2.48)	4.42 (2.59)	0.516 (0.03)			
Hip frontal RoM [°]	12.48 (2.97)	11.80 (2.68)	11.79 (3.17)	0.096 (0.11)			
Hip peak abduction moment [Nm/kg]	0.19 (0.08)	0.19 (0.10)	0.21 (0.09)	0.055†			
Hip peak adduction moment [Nm/kg]	-0.92 (0.10)	-0.95 (0.14)	-0.94 (0.17)	0.614 (0.02)			
Pelvic obliquity RoM [°]	6.31 (2.16)	6.68 (2.02)	6.59 (1.91)	0.329 (0.05)			
Hip peak int. rotation [°]	-6.71 (10.84)	-6.35 (10.25)	-4.50 (10.27)	0.172†			
Hip peak ext. rotation [°]	5.47 (10.34)	5.88 (9.62)	9.37 (10.76)	0.077 (0.12)			
Hip transverse RoM [°]	12.18 (3.08)	12.23 (3.25)	13.87 (4.49)	0.023 (0.17)	0.937	0.069	0.069
Hip peak internal rotation moment [Nm/kg]	-0.07 (0.04)	-0.07 (0.03)	-0.07 (0.03)	0.565†			
Hip peak external rotation moment [Nm/kg]	0.20 (0.05)	0.21 (0.06)	0.22 (0.07)	0.280 (0.06)			
Pelvic rotation RoM [°]	10.41 (4.35)	11.37 (3.95)	12.49 (5.41)	0.001†	0.026	0.001	0.026

Note. P-values as revealed by one-way repeated measures ANOVAs / Friedman tests and Holm-Bonferroni corrected pairwise comparisons and effect sizes (η_p^2); significant results ($p < 0.05$) in bold; † marks analysis with a Friedman test. B = baseline; S = short-term; M = medium-term; RoM = range of motion.

8.5 Discussion

The present study first aimed to quantify the effects of moderate unilateral HOA on gait biomechanics, proprioception and functional capacity. The main findings are that, in line with our hypotheses, individuals with HOA walked significantly more slowly with a reduced step length, sagittal hip RoM and peak extension angle and had a reduced functional capacity. In contrast to our expectations, no effect of HOA on hip proprioception was found.

The second aim of this study was to evaluate the short- and medium-term effects of a hip brace on gait biomechanics, hip proprioception, pain perception and functional capacity. In summary, as hypothesised, after medium-term brace application, step length, walking speed and functional capacity were significantly increased. Additionally, pain perception was significantly lower in the intervention period. However, no brace effects on hip sagittal RoM, peak extension angle or hip proprioception were found, neither after short-term nor after medium-term brace application.

8.5.1 Effects of hip osteoarthritis on functional capacity, hip proprioception and gait biomechanics

The effects of HOA on gait biomechanics observed in this study were alterations of the gait pattern that have been previously described (e.g. Constantinou et al., 2017; Watelain et al., 2001). Thus, the present HOA group presented typical symptoms of HOA gait. Walking speed of individuals with HOA was significantly lower than that of CON, with a mean walking speed of 1.23 m/s, representing a difference of 6.8 %. This is a smaller difference than reported by Constantinou et al. (2014) who, in a meta-analysis, calculated a mean walking speed of 0.95 m/s and a mean difference of 26 % compared to the CON. However, individuals with end-stage HOA scheduled to undergo THR were included in their analysis. The authors attributed the reduction in gait speed to a reduction in step length of the affected limb (Constantinou et al., 2014), which was also observed in the present study. Additionally, individuals in the present study showed limited mobility in hip extension and hip transverse rotation. The limitations in dynamic RoM have previously been attributed either to limitations in passive joint mobility (Baker et al., 2016; Eitzen et al., 2012; Holla et al., 2011) or to a strategy to reduce joint loading and pain (Hurwitz et al., 1997; Meyer et al., 2018; Ornetti et al., 2011).

For both subject groups, peak sagittal hip angles observed in this study deviated from those reported previously by e.g. Eitzen et al. (2012), Ornetti et al. (2011) or Tateuchi et al. (2014). While the RoM of 45° from the CON group is comparable to the data presented by these authors, the peak values are shifted towards lower hip flexion and more pronounced hip extension. This offset in sagittal hip rotation was previously described by authors comparing different biomechanical models and is likely to be caused by differences in the definition of pelvis neutral position (Falisse et al., 2018; Roelker et al., 2017). Consequently, deviations in hip joint angles must be interpreted with caution when comparing different studies, and individual model definitions have to be considered in this context.

The limitations observed for hip joint excursion were reflected in reduced external hip flexion, extension, adduction and external rotation moments. This is in line with data from Tateuchi et al. (2021) who reported a reduction of all peak joint moments with decreased step length. The reduction in peak adduction moment has also been reported in a recent meta-analysis of HOA walking dynamics (Diamond et al., 2018). However, the authors suggested that the alterations only occur in individuals with end-stage HOA. One reason for this might be that the studies included in the meta-analysis used matched walking speeds or statistical methods to correct for different walking speeds between the healthy and the HOA groups. As hip adduction moment correlates with walking speed (Rutherford & Hubley-Kozey, 2009) this might artificially lower the hip adduction moment observed for healthy controls. However, during natural everyday walking conditions, joint moments are lower even in individuals with mild-to-moderate HOA, as seen in this study. Thereby, the peak adduction moment is thought to be a key parameter for hip joint loading (Wesseling et al., 2015) as it has to be counteracted by the hip abductor muscles. Tateuchi et al. (2017) reported that the daily cumulative hip moment in the frontal plane, in particular, and potentially the cumulative hip moment in the sagittal plane, are predictors of radiographic progression of HOA. Thus, reduced hip joint moments are likely to originate in pain avoidance gait strategies adopted by the affected individuals and decreased muscular function (Meyer et al., 2018; Ornetti et al., 2011; Watelain et al., 2001).

Besides the changes in hip joint biomechanics, the RoM of the pelvis in the sagittal plane (pelvis tilt) was significantly increased. Similar results were found by Watelain et al. (2001) in a group of people with early-stage HOA. Lee et al. (1997) stated that “anterior pelvic tilting was the most closely associated compensatory mechanism for reduced hip extension during gait”. In contrast, pelvis motion was not a predominant gait feature of HOA gait in a study by Meyer et al. (2015).

The decreased walking speed and step length resulted in a significant reduction of functional capacity assessed using the 6MWT. Despite a comparable HHS, the distances covered during this study by either the HOA or the control group were lower than those reported by Eitzen et al. (2015) and Rydevik et al. (2010) who reported distances between 630 m and 673 m for individuals with HOA and 719 m for healthy controls. One explanation for this might be different test settings with a circuit of 54 m length in our study and a corridor of 20 m length in the studies of Eitzen et al. (2015) and Rydevik et al. (2010). Likewise, Kumar et al. (2015) reported a value of 628 m for individuals with mild-to-moderate HOA. They, however, defined HOA solely based on radiographic imaging and did not include functional criteria. The reported

HOOS values for pain and activities of daily living were higher than in the present study, indicating less pain and functional impairments.

To our knowledge, this was the first study to quantify hip proprioception in a cohort of individuals with HOA. The applied angle reproduction test did not reveal any significant effects of HOA on sagittal plane hip proprioception. This result may indicate that, in contrast to KOA, HOA does not cause a loss in proprioceptive perception, or that the measurement technique applied in this study was unable to detect any differences. Pickard et al. (2003) compared the hip joint position sense of younger and older adults using an angle reproduction test in the frontal plane, and likewise did not find any significant differences between the groups. Thus, in contrast to knee joint position sense, for which a decline with age has frequently been reported (Hurley et al., 1998; Kaplan et al., 1985), no age effect was detectable for the hip joint. Hillier et al. (2015) stated that active angle reproduction tests, while reflecting the functional use of proprioception, require sufficient kinaesthetic memory of the pre-established position as well as motor control to readopt the joint position. As hip flexion movement was performed in a one-legged standing position, maintaining balance as well as flexing the hip joint against gravity may have required a large portion of motor control. Therefore, functional limitations in this area might have masked underlying differences in the sensory quality of hip proprioception. Performing the hip flexion angle reproduction test in a supine position may eliminate this interference.

8.5.2 Effects of hip bracing on pain, functional capacity, hip proprioception and gait biomechanics

The application of a hip brace had two aims: (I) stabilising the pelvis and stimulating several trigger points at the GMed and SIJ, and (II) mobilising the hip joint by applying a friction massage to the muscle insertions at the greater trochanter. Application of the brace resulted in a significant reduction of the perceived pain during walking activities and at night. Thereby, 18 out of 21 participants reported a pain reduction during walking activities, and 14 out of 21 participants reported a decrease in night pain. In contrast, brace application in the study of Nérot & Nicholls (2017) only resulted in pain reduction in 9 of 14 participants. They, however, only analysed immediate pain relief. In a longitudinal study on the effects of the WISH type of S-form hip brace, pain perception decreased the most in the first three months of brace use (T. Sato et al., 2008). Thus, longer brace application might further increase the positive effects.

The observed absolute changes in VAS during this study of 10.5 mm (participants with less pain) and 11.4 mm (participants with more pain) for walking activities, as well as 6.2 mm (participants with less pain) and 3.1 mm (participants with more pain) for night pain, are rather small. A study aiming to determine the minimal clinically important difference (MCID) in a cohort of individuals with HOA reports values of 15.3 mm or 32 % (Tubach et al., 2005). Thus, in absolute values, the clinical relevance of the shown improvements or deteriorations is questionable. However, the relative changes (less pain: 45.5 % & more pain: 40.8 % for walking activities; less pain: 41.6 % & more pain: 84.2 % for night pain) exceed the proposed MCID of 32 %. Hence, brace-induced changes in pain perception might still be perceived as beneficial or detrimental. Additionally, Dekker (2019) states that anchor-based methods to detect MCIDs yield highly variable results. The author further stresses that the MCID is highly subjective and that the individual patient must decide which improvement is important enough to undergo treatment. Further analyses of gait biomechanics at an individual level might provide additional insights on why, in a small portion of participants, brace application leads to an increase in pain.

Besides the changes in pain perception, participants were able to increase their step and stride length as well as walking speed, which resulted in an increased performance during the 6MWT. The distance covered during the six minutes increased on average by 5 %. In a study with people with KOA, it was shown that performance in the 6MWT was highly correlated with the KOA outcome score subscales for pain and quality of life (Ateef et al., 2016). Despite the increase in walking distance covered during the 6MWT, the pain level after the task remained unchanged. Thus, functional capacity increased without negative effects on pain perception. E. Sato et al. (2012) analysed hip brace effects using the TUG and, similarly, found increases in functional capacity. Furthermore, the improvement increased after long-term brace use. Thus, long-term brace usage might further increase the positive effects on functional capacity observed after medium-term brace application.

In contrast to our expectations, no effect of bracing on hip proprioception was observed. While this contradicts some findings published for knee bandages (Baltaci et al., 2011; Beynon et al., 1999, 2002; Bodendorfer et al., 2019; Selfe et al., 2008, 2011), it is not surprising, given that no reduction in joint position sense was found for the HOA group. However, positive brace effects have also been found in healthy young people who should not exhibit proprioceptive loss (Baltaci et al., 2011). Therefore, again, the testing procedure might not be sensitive enough to detect smaller changes in hip joint position sense.

Brace application overall did not lead to a “normalisation” of the gait patterns in that typical symptoms of HOA (e.g. decreased hip extension or reduced sagittal RoM) remain even after medium-term brace application. This finding aligns with results from patients treated with THR. Beaulieu et al. (2010) showed that alterations in gait biomechanics such as reduced peak flexion angle, peak extension angle and sagittal plane RoM remained even ten months after surgery. Similar results were found by Foucher et al. (2007) as well as Zügner et al. (2018) even one and two years after THR, respectively. Beaulieu et al. (2010) and Foucher et al. (2007) speculated that pain-avoidance strategies adopted pre-operatively might persist even after successful surgery, or that alterations are caused by persistent muscle weaknesses. The one-week intervention period might therefore not have been sufficient time for the participants to adopt a normal gait pattern.

Brace application, in contrast, induced some additional biomechanical changes. Short-term brace application resulted in a significant reduction of the peak hip flexion angle, and there is a tendency for this effect to remain even after medium-term brace application. This reduction in the peak flexion angle may reflect the passive resistance of the brace to hip flexion movement. Nérot & Nicholls (2017) reported that brace application led to a feeling of either restriction or support by the participants. Unfortunately, no kinematic data on hip flexion were presented.

Brace application not only influenced hip but also pelvis kinematics. In both braced conditions, pelvic tilt and rotation RoM were increased in comparison with the unbraced condition. Anterior tilting of the pelvis, as well as more pronounced rotation of the pelvis, might allow the patients to increase step length despite limitations in hip extension and internal rotation mobility (Leigh et al., 2016). Liang et al. (2014) demonstrated that for gait velocities above 1.0 m/s pelvic rotation lengthens the step, which is accompanied by an increase in pelvic rotation RoM.

As the pelvis is closely connected to the lumbar spine (Ike et al., 2018; Thurston A J & Harris, 1983; Whittle & Levine, 1999), alterations in pelvis motion are likely to affect lumbar spine mobility (Hurwitz et al., 1997; Watelain et al., 2001). People with HOA frequently report lower back pain (Thurston, 1985). Previous studies on lower back pain have shown an association between pain and increased pelvic horizontal rotation (Huang et al., 2011). In contrast, people with acute lower back pain exhibit reduced pelvic rotation, which may help to reduce forces transmitted across the lumbar spine (N. Taylor et al., 2004). Thus, increased pelvic rotation without according lumbar spine movement might increase stress on the lumbar spine and the SIJ (Watelain et al., 2001). However, increased pelvic and lumbar spine

motion is not correlated with back pain in patients with HOA (Thurston, 1985). The applied hip brace exerts pressure on the pelvis and onto several trigger points at the SIJ. Future research should clarify whether increases in pelvic motion observed with bracing are enabled by a stabilisation of the pelvis-spine complex and/or pain reduction, and whether increased pelvic motion causes greater stress on the lumbar spine, potentially triggering lower back pain with long-term brace usage.

The observed kinematic alterations were reflected in an increase in the external peak extension moment for both braced conditions in comparison to the unbraced condition. Tateuchi et al. (2021) reported an increase for all peak joint moments when gait velocity increased, especially when velocity increase was achieved by an increase in step length rather than cadence. However, despite an increase in gait velocity and step length, none of the other hip joint moments increased in this study. The peak of the extension moment is reached during terminal stance. A longer step length caused by a more forward tilted pelvis might increase the distance between the foot (and hence the centre of pressure) and the CoM (Leigh et al., 2016). Thus, the GRF lever arm might be increased, resulting in larger extension moments. Similarly, in a longitudinal study investigating the effects of a medical foot device in a population of people with HOA over one year, an increase in hip extension moment accompanied by a decrease in hip pain was found (Solomonow-Avnon et al., 2017). This is in line with results from Hurwitz et al. (1997) who reported a negative correlation between hip pain and peak hip extension moment. Thus, the pain relief caused by brace application might have enabled the HOA patients to increase the hip extension moment. Furthermore, application of the WISH-type hip brace resulted in larger vertical GRF during walking (Yamaji et al., 2009). Although no joint moments were reported, they are likely to increase with increasing GRF (Toda et al., 2015).

In summary, this study demonstrated a positive effect of hip brace application on hip pain and functional capacity in a cohort of people with mild-to-moderate unilateral HOA. Thereby, patients in our study demonstrated typical biomechanical features of HOA gait. Brace application increased walking speed, step length and external peak extension moment. Additionally, brace application impacted pelvis mobility with increased RoM in the sagittal and transverse planes. As the applied brace does not intend to mechanically alter hip joint biomechanics, but rather influence the soft tissue structures surrounding the hip joint, both immediate and medium-term effects were observed. Lastly, it was found that neither HOA nor hip bracing altered hip proprioception in the sagittal plane.

8.5.3 Limitations

Alongside the strengths of our study, there are also some limitations.

(I) The order of the bracing conditions in the HOA group was not randomised, raising the risk for potential sequencing effects. Mild to moderate HOA is characterised by alternating phases of less and more pain (Bastick et al., 2016). Thus, to minimise the effects of pain fluctuation on the study results, we intended to have brief intervals between the test sessions. As the primary focus of the hip brace is manipulation of soft tissue, application might cause long-term effects beyond the period of application, and an according wash-out period between test sessions would have been necessary. Therefore, randomisation of bracing conditions was not conducted. Additionally, it has to be noted that due to the duration of brace production and availability of the participants, the time interval between session one and session two was substantially longer (5 ± 4 weeks) than the time interval between session two and session three (1 ± 0 week).

(II) Equally, for obvious reasons, it was not possible to blind the participants to the brace intervention, which enables a potential placebo effect, especially on subjective parameters such as pain perception.

(III) For the biomechanical gait analysis, participants were equipped with retroreflective markers on the pelvis. In sessions two and three, participants wore the brace during the measurements, and thus, the ASIS and PSIS were covered by the pelvis belt of the brace. Hence, markers on the ASIS had to be placed on the hip belt while markers on the PSIS were attached through the mesh fabric of the pelvis belt. Application of markers on clothing increases the risk of relative movement between the bony segment and the marker (Milner, 2008). The application of other methods, such as clusters, often reported for knee brace analyses (Focke et al., 2020; Turner et al., 2021), is not applicable for the hip joint, as the pelvis was entirely covered by the brace, and adjoining segments (thighs and torso) possess large portions of wobbling mass. However, as the pelvis belt of the brace was fitted very tightly, relative movement between the pelvis and the brace is probably small but cannot be fully excluded.

(IV) The results of our study confirmed the often-reported effect of reduced gait velocity in patients with HOA. Additionally, gait velocity increased after medium-term brace application. Thus, gait velocity differed between the CON and the HOA group as well as within the HOA group under different bracing conditions. It has previously been shown that gait velocity correlates with hip joint kinematics and dynamics (Fukuchi et al., 2019; Lelas et al., 2003).

Therefore, observed changes in gait biomechanics might reflect differences in gait velocity to some degree. Several methods, including prescribed walking speeds or statistical methods (e.g. the analysis of covariance (ANCOVA)), have been used to account for such different walking speeds (Astephens Wilson, 2012). However, the use of prescribed gait speeds contains the risk of not capturing normal gait patterns as people are forced to walk with a prescribed speed and thus does not reflect, e.g. joint loading on a daily basis (Astephens Wilson, 2012). Additionally, it has been shown that gait modifications in people with HOA persist even at matched gait speeds (Ismailidis et al., 2021). The application of ANCOVA neglects the fact that differences in gait speeds between groups are not a source of random error variability but rather representative of the population characteristics (Astephens Wilson, 2012). Therefore, no prescriptions or corrections regarding gait velocity were applied in this study as previously applied by others (Eitzen et al., 2012; Meyer et al., 2018).

8.5.4 Outlook

This was one of the first studies to present comprehensive data regarding the effects of hip bracing for patients with HOA on hip and pelvis biomechanics, proprioception, functional capacity and pain perception. Thereby, not only immediate but also medium-term effects were presented. Brace application resulted in several alterations of the gait biomechanics, functional capacity and pain perception. However, HOA as well as hip braces have been shown to impact the movement and loading at other body locations such as the knee (Hurwitz et al., 1997; Nérot & Nicholls, 2017; Rutherford et al., 2015b; Tateuchi et al., 2014), ankle (Kubota et al., 2007; Ornetti et al., 2011; Schmitt et al., 2015) and lumbar spine (Murray et al., 1971; Thurston, 1985). Alterations occurring at the hip joint might therefore impact the function of other body locations and might trigger or prevent concomitant diseases such as KOA (Rutherford et al., 2015b; Shakoor et al., 2014; van Drongelen, Wesseling, et al., 2020). Therefore, future research should include analyses of the contralateral limb, adjacent joints or even whole-body movement. Thereby, the application of pattern recognition methods such as principal component analysis or cluster analysis on whole body kinematics, as proposed by Meyer et al. (2015), van Drongelen, Stetter, et al. (2021) and Stetter et al. (2020), might allow us to gain additional insights into HOA and brace effects in other joints or the entire body.

While the biomechanical effects of hip brace application remain partly unexplained, positive effects on pain perception and functional capacity are encouraging enough to expand the research on hip bracing for patients with mild-to-moderate HOA. Thereby, special attention should be paid to the long-term effects of brace application. As the brace used in the present study is not designed to mechanically alter hip joint movement but instead focuses on soft tissue manipulation, longer periods of brace application might further increase the effects already observed after one week. Therefore, analyses of brace effects on muscle activity during gait should be included, as people with HOA have been shown to have decreased muscle strength (Loureiro et al., 2013).

8.6 Conclusion

The present study is one of the first comprehensive studies to quantify the kinematic, dynamic, proprioceptive and functional effects of a hip brace in a clearly defined cohort of individuals with mild-to-moderate unilateral HOA. Additionally, insights on the effects of HOA on gait biomechanics, hip proprioception and functional capacity were gained or extended. While hip proprioception was not influenced by either HOA or hip bracing, there were substantial impacts of both parameters on functional capacity and gait biomechanics. Brace application resulted in reduced pain perception and higher functional capacity. However, as key parameters of HOA gait, such as a reduced peak extension angle, remained unchanged, the underlying mechanisms remain partly unclear. Future studies should therefore include additional data regarding whole-body biomechanics or muscle activation and extend the brace application period to gain insights into long-term effects of bracing.

9 **Manuscript VI - Effect of hip osteoarthritis and hip bracing on turning biomechanics**

Slightly modified version of the paper published as:

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9.1 Abstract

Turning movements are frequently encountered during daily life and require more frontal and transverse hip mobility than straight walking. Thus, analysis of turning might be an insightful addition in the evaluation of conservative treatment approaches for HOA patients. The study objective was to quantify the effects of mild-to-moderate symptomatic HOA on lower-body turning biomechanics and evaluate the effects of hip bracing in this cohort.

Biomechanical analysis of 90° step and spin turns in 21 persons with HOA and 21 healthy controls (case-control study) and intervention study on the effects of hip bracing in HOA participants. Hip and knee kinematics and dynamics were compared using independent sample t-tests or one-way repeated measure ANOVAs.

Persons with HOA have reduced peak hip extension and sagittal and transverse hip RoM during turning. During the spin turn, reduced hip adduction and frontal hip RoM were found. Bracing increased the movement velocity at turn initiation and limited the transverse hip RoM during both turns, but increased knee peak adduction and internal rotation moments during spin turns.

Persons with HOA present altered hip kinematics during 90° spin turns in all movement planes. Their inclusion during clinical movement analysis might facilitate the detection of mobility deficits at an early disease stage. Bracing led to higher movement velocity at turn

initiation without increasing load at the hip joint and reduced transverse hip RoM, avoiding the painful reaching of RoM endpoints.

9.2 Introduction

People with HOA suffer from pain and loss of function, leading to a decrease in quality of life (Salaffi et al., 2005). Studies have shown that half of the persons with HOA undergo hip replacement surgery within 2.5 years after initial consultation (Dabare et al., 2017), and hip replacement rates increased by 22 % between 2009 and 2019 (OECD, 2021). If this trend continues, the economic burden might become unsustainable (Ackerman et al., 2019). Therefore, there is an urgent need for effective conservative treatment strategies to reduce pain and enhance quality of life to avoid or postpone hip surgery. One conservative treatment option is the use of braces, which has been found to decrease hip pain and improve function (E. Sato et al., 2012; T. Sato et al., 2008; Steingrebe et al., 2022). Thereby, increased overall Harris Hip Scores (HHS) of 21.6 points or 37 % (T. Sato et al., 2008) and reduced time taken for the TUG (E. Sato et al., 2012) were reported after one month of using the rigid WISH-type hip brace. Reduced pain during walking and reduced night pain by about 28 % and 18 %, respectively, and a 5 % increase in the distance covered during the 6MWT were found after one week of using the elastic CoxaTrain hip brace. Additionally, biomechanical analyses of straight walking showed a significant increase in gait speed and step length (Steingrebe et al., 2022). However, the underlying mechanisms of these benefits remain unclear. Analysis of straight walking has shown the effects of bracing on pelvic tilt and pelvic rotation RoM, but no effects on hip joint loading or frontal or transverse hip kinematics were found (Steingrebe et al., 2022). However, loading and joint mobility requirements in the frontal and transverse planes might be larger during other movements of daily living, such as turning. Turning movements are encountered frequently during activities of daily living (Boekesteijn et al., 2023; Glaister et al., 2007) and are a risk factor associated with falls (Leach et al., 2018). Turning can be conducted either in a spin or step turn manner (Hase & Stein, 1999): compared to straight walking, the step turn requires larger hip external rotation around terminal stance and abduction throughout the ST, and the spin turn requires larger hip external rotation throughout the ST and adduction around mid stance (M. M. Taylor et al., 2005). People with HOA have demonstrated decreased peak hip abduction and adduction during 45° turns (Tateuchi et al., 2014). The turning angle has been found to impact lower body joint moments in the frontal and transverse planes (Xu et al., 2006) and previous research has shown that the majority of turns are between 76° and 120° (Sedgman et al., 1994). Therefore, analysing

turns in this angle range is more relevant for activities of daily living. Additionally, in previous studies, analyses of transverse plane biomechanics are lacking.

Therefore, the first aim of this study was to quantify the effect of unilateral mild-to-moderate HOA on 3D hip and knee biomechanics during 90° turns while walking. We hypothesised that persons with HOA would demonstrate reduced hip adduction during a spin turn, reduced hip abduction during a step turn and reduced peak hip extension during both turning tasks. The second aim was to identify changes in turning biomechanics imposed by short- and medium-term application of a hip brace in a cohort of persons with mild-to-moderate unilateral HOA. We expected bracing to modify hip biomechanics, especially in the frontal and transverse planes, during both turning conditions.

9.3 Materials and Methods

9.3.1 Participants

The study was conducted in a cohort of 21 persons with mild-to-moderate unilateral primary HOA (10 females; age: 64.0 ± 9.6 years; BMI 24.2 ± 2.9) and 21 healthy control participants (10 females; age: 63.1 ± 9.2 years; BMI 25.2 ± 2.7) matched to the mean age, weight and height of the HOA group (Table 7.1). Hip osteoarthritis was defined radiologically and functionally: Participants from the HOA group had an ipsilateral K-L grade of two or above on a standing anterior-posterior X-ray, evaluated by the same experienced orthopaedist, and showed mild-to-moderately decreased hip function based on an HHS between 65 and 95. Detailed inclusion and exclusion criteria can be found in Table 6.1. The study protocol was approved by the ethical committee of the KIT and conforms to the Declaration of Helsinki. All participants gave their written informed consent before study participation.

9.3.2 Testing protocol

All measurements were recorded from April 2019 until March 2021 at the same laboratory. Participants with HOA presented at the lab on three occasions: in the first session, baseline measurements without bracing were conducted, in the second session, participants were individually fitted with a hip brace (Figure 8.2; Coxatrain®, Bauerfeind AG, Zeulenroda-Triebes, Germany) and tested after a few minutes of brace familiarisation (short-term brace effects). The third session was conducted after one week of brace wear (> 4 hours per day)

during everyday activities (medium-term brace effects). Participants of the control (CON) group were tested on one occasion without bracing. The testing protocol was equal in all sessions: participants performed pre-planned 90° step and spin turns (Figure 9.1) while walking at a self-selected speed. Approach speed was controlled using light barriers and kept constant (within $\pm 5\%$ of the first trial) across trials of the same turning condition within the same session. For each turning condition, five valid trials were recorded. Participants were allowed to initiate the turn within a marked corridor of 50 cm and instructed to perform the turn in an abrupt manner. Turning was always conducted with the affected limb for the HOA participants and a matched limb for the control participants.

9.3.3 Hip brace

The hip brace (Figure 8.2) applied in this study consists of an elastic pelvis and thigh belt connected by a rigid hinge joint with one degree of freedom (flexion/extension). The brace intends to stabilise the pelvis and stimulate trigger points around the GMed and the SIJ using viscoelastic pads within the bandage. Additionally, a pad positioned above the greater trochanter moves vertically with extension/flexion of the brace joint and applies a friction massage to the muscle insertions at the greater trochanter.

9.3.4 Data acquisition and processing

To assess hip and knee biomechanics, 3D motion analysis was conducted using a 16-camera infrared motion capturing system (200 Hz, Vicon Motion Systems, Oxford Metrics Group, Oxford, United Kingdom) and two 3D force plates (1000 Hz, BP600900, Advanced Mechanical Technology Inc., Watertown, MA, United States). A full-body marker set with 42 retroreflective markers was used, and markers were applied by the same experienced researcher for all participants (Härtel & Hermsdorf, 2006). Kinematic and GRF data were filtered using a 4th-order Butterworth low-pass filter with a cut-off frequency of 15 Hz. An inverse kinematics and dynamics approach using the multi-body model ALASKA Dynamicus (Härtel & Hermsdorf, 2006), including the HJC definition proposed by Harrington et al. (Harrington et al., 2007; Kainz et al., 2015), was used to calculate 3D joint angles and external joint moments. Peak joint angles and RoM, as well as peak external joint moments of the hip and knee during the ST of turning, were assessed. Additionally, the foot-progression angle (FPA) at foot flat (positive values indicate outward rotation of the foot relative to the approach axis; Figure 9.1), the forward-directed CoM velocity at IC and TO and the ST duration were assessed.

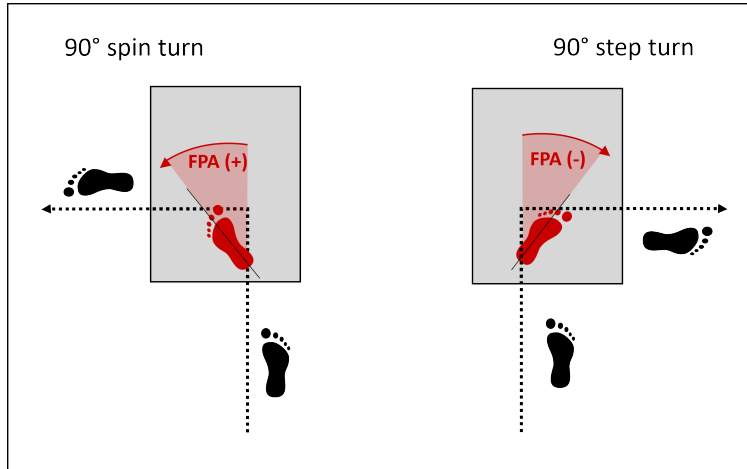


Figure 9.1: Schematic representation of foot placement during 90° spin turn (left) and 90° step turn (right).
Note. Red feet = affected limbs in hip osteoarthritis participants and matched limbs in healthy controls. Calculation of foot progression angle (FPA) between approach axis and longitudinal foot axis with (+) outward rotation and (-) inward rotation of the foot.

9.3.5 Statistical analysis

The sample size was calculated *a priori* based on the study of Nérot & Nicholls (2017) on the effects of bracing on hip joint kinematics during straight walking in 14 participants with unilateral HOA using G*Power (version 3.1.9.3) (Faul et al., 2007). The observed effect sizes in their study were 0.92 and 1.17 for peak hip adduction and internal rotation, respectively. With a significance criterion of $\alpha = 0.05$ and power = 0.95, the minimum sample sizes required were 18 and 12, respectively.

All statistical analyses were performed in R (version 4.2.2). Normal distribution and homogeneity of variance were assessed using the Shapiro-Wilk and Levene tests, respectively. Differences between the HOA and CON groups at baseline were analysed using independent sample t-tests. To assess the impact of HOA severity, parameters with significant group differences were subsequently correlated with the HHS within the HOA group using Spearman's rank correlation. Brace effects within the HOA group were assessed using univariate repeated-measures ANOVAs. If sphericity was violated, Greenhouse-Geisser estimates were used. Post hoc analysis was conducted using t-tests for dependent samples with Holm-Bonferroni corrections. If the conditions for parametric testing were not fulfilled, nonparamet-

ric tests (MWU, Friedman or WCX) were used. The level of significance was set *a priori* to $\alpha < 0.05$. Cohen's *d* (Cohen, 1988) and partial eta squared were used to calculate effect sizes for between-group and within-group comparisons, respectively.

9.4 Results

Hip and knee joint angle time curves for the spin and step turn can be found in the Appendix (Figure A.28 to Figure A.31).

9.4.1 Effects of hip osteoarthritis on hip and knee biomechanics

All results on the comparison of hip and knee biomechanics during the step and spin turn between the HOA group at baseline and the CON group are presented in Table 9.1 and Table 9.2, respectively. During the step turn, HOA participants demonstrated reduced peak hip extension (mean difference (MD) = -4.82° ; $p = 0.01$; $|d| = 0.84$) and decreased sagittal (MD = 6.27° ; $p = 0.005$; $|d| = 0.91$) and transverse (MD = 9.96° ; $p < 0.001$; $|d| = 1.43$) hip RoM. For the knee joint, increased peak flexion (MD = -3.70° ; $p = 0.02$; $|d| = 0.78$) was found.

During the spin turn, reduced peak hip extension (MD = -4.64° ; $p = 0.01$; $|d| = 0.84$) and adduction (MD = -1.99° ; $p = 0.02$; $|d| = 0.75$), as well as reduced hip RoM in all movement planes (sagittal: MD = 5.10° ; $p = 0.008$; $|d| = 0.86$; frontal: MD = 1.76° ; $p = 0.03$; $|d| = 0.71$; transverse: MD = 6.01° ; $p = 0.03$; $|d| = 0.70$), were found in the HOA group. No differences in knee biomechanics were observed.

All results from the correlation analyses assessing the impact of HOA severity on turning biomechanics can be found in the Appendix (Table A.8 and Table A.9). Of the analysed parameters, the transverse hip RoM during the step turn showed a moderate positive correlation with the HHS ($\rho = 0.36$; $p = 0.112$) while the peak hip extension angle during the spin turn showed a negative moderate correlation with the HHS ($\rho = -0.36$; $p = 0.106$). This means that larger hip transverse RoM during the step turn and larger hip extension angles during the spin turn are associated with better hip function based on the HHS.

Table 9.1: Results of discrete hip and knee joint angle and external joint moment parameters during the step turn for the hip osteoarthritis (HOA) group at baseline and the control group.

Parameter during step turn		HOA Base-line	Control Group	Mean difference	t-test/MWU
		Mean (SD)	Mean (SD)	(95%CI)	p (d)
Sagittal plane	Hip peak extension [°]	-20.32 (5.73)	-25.13 (5.76)	-4.82 (-8.40 -1.24)	0.010 (0.84)
	Hip peak flexion [°]	8.18 (6.84)	9.64 (5.58)	1.46 (-2.44 5.35)	0.454 (0.23)
	Hip sagittal RoM [°]	28.50 (8.01)	34.77 (5.58)	6.27 (1.97 10.58)	0.005 (0.91)
	Hip peak extension moment [Nm/kg]	2.82 (2.80)	2.61 (2.74)	0.13 (-0.90 1.16)	0.746 (0.08)
	Hip peak flexion moment [Nm/kg]	-2.77 (1.80)	-2.63 (1.48)	0.21 (-1.52 1.94)	0.797 (0.08)
	Knee peak extension [°]	11.05 (4.165)	8.88 (3.218)	-2.17 (-4.49 0.15)	0.066 (0.58)
	Knee peak flexion [°]	44.37 (4.74)	40.67 (4.71)	-3.70 (-6.64 -0.75)	0.015 (0.78)
	Knee sagittal RoM [°]	33.32 (5.25)	31.79 (4.02)	-1.53 (-4.44 1.39)	0.295 (0.33)
	Knee peak extension moment [Nm/kg]	-2.37 (1.59)	-2.21 (1.42)	0.17 (-0.78 1.11)	0.723 (0.11)
	Knee peak flexion moment [Nm/kg]	0.72 (1.22)	0.75 (1.20)	0.03 (-0.72 0.79)	0.517 (0.03)
Frontal plane	Hip peak adduction [°]	-3.26 (2.53)	-2.61 (2.76)	0.65 (-1.00 2.30)	0.433 (0.24)
	Hip peak abduction [°]	5.47 (2.98)	6.71 (3.37)	1.24 (-0.75 3.22)	0.215 (0.39)
	Hip frontal RoM [°]	8.73 (2.39)	9.32 (2.41)	0.59 (-0.91 2.09)	0.430 (0.25)
	Hip peak adduction moment [Nm/kg]	-1.25 (1.47)	-1.29 (1.75)	-0.04 (-1.05 0.97)	0.369 (0.03)
	Hip peak abduction moment [Nm/kg]	3.50 (3.01)	3.89 (3.13)	0.40 (-1.52 2.31)	0.396 (0.13)

Table 9.1: Results of discrete hip and knee joint angle and external joint moment parameters during the step turn for the hip osteoarthritis (HOA) group at baseline and the control group.

Parameter during step turn		HOA Base-line	Control Group	Mean difference	t-test/MWU
		Mean (SD)	Mean (SD)	(95%CI)	p (d)
Frontal plane	Knee peak adduction [°]	-2.66 (3.66)	-3.62 (4.56)	-0.96 (-3.54 1.62)	0.862 (0.23)
	Knee peak abduction [°]	2.36 (4.18)	1.45 (4.27)	-0.92 (-3.55 1.71)	0.485 (0.22)
	Knee frontal RoM [°]	5.03 (2.51)	5.07 (2.41)	0.04 (-1.49 1.58)	0.921 (0.02)
	Knee peak adduction moment [Nm/kg]	3.56 (3.59)	3.84 (3.77)	0.29 (-2.01 2.58)	0.584 (0.08)
	Knee peak abduction moment [Nm/kg]	-1.81 (2.08)	-1.80 (2.18)	0.01 (-1.32 1.34)	0.709 (0.01)
Transverse plane	Hip peak int. rotation [°]	-13.82 (13.68)	-23.07 (16.03)	-9.2 (-18.55 0.03)	0.051 (0.62)
	Hip peak ext. rotation [°]	6.97 (11.45)	7.68 (14.33)	0.70 (-7.38 8.79)	0.861 (0.05)
	Hip transverse RoM [°]	20.79 (7.85)	30.75 (5.97)	9.96 (5.61 14.31)	<0.001 (1.43)
	Hip peak internal rotation moment [Nm/kg]	-1.81 (0.81)	-2.21 (0.94)	-0.41 (-0.95 0.14)	0.141 (0.46)
	Hip peak external rotation moment [Nm/kg]	0.58 (0.61)	0.46 (0.52)	-0.12 (-0.48 0.23)	0.330 (0.22)
	Knee peak int. rotation [°]	28.46 (11.62)	32.71 (13.69)	4.25 (-3.67 12.17)	0.285 (0.33)
	Knee peak ext. rotation [°]	42.57 (12.65)	45.49 (13.61)	2.93 (-5.27 11.12)	0.356 (0.22)
	Knee transverse RoM [°]	14.11 (3.97)	12.78 (2.87)	-1.32 (-3.49 0.84)	0.223 (0.38)
	Knee peak internal rotation moment [Nm/kg]	0.95 (0.49)	0.76 (0.37)	0.76 (-0.46 0.08)	0.164 (0.44)
	Knee peak external rotation moment [Nm/kg]	-1.30 (0.84)	-1.47 (0.99)	-0.18 (-0.75 0.39)	0.485 (0.19)

Table 9.1: Results of discrete hip and knee joint angle and external joint moment parameters during the step turn for the hip osteoarthritis (HOA) group at baseline and the control group.

Parameter during step turn		HOA Base-line	Control Group	Mean difference	t-test/MWU
		Mean (SD)	Mean (SD)	(95%CI)	p (d)
Spatio-temporal	Stance phase duration [s]	0.73 (0.10)	0.70 (0.08)	-0.03 (-0.08 0.02)	0.230 (0.34)
	Forward CoM velocity at IC [m/s]	0.94 (0.14)	0.97 (0.12)	0.03 (-0.05 0.11)	0.431 (0.25)
	Forward CoM velocity at TO [m/s]	0.94 (0.16)	0.98 (0.15)	0.04 (-0.06 0.14)	0.398 (0.26)
	Foot progression angle at foot-flat	-37.05 (15.88)	-34.68 (18.83)	2.37 (-8.49 13.24)	0.661 (0.14)

Note. P-values as revealed by independent sample t-tests/Mann-Whitney-U tests (MWU) and effect sizes (|d|); significant results ($p < 0.05$) in bold. RoM = range of motion; CoM = centre of mass; IC = initial contact; TO = toe off.

Table 9.2: Results of discrete hip and knee joint angle and external joint moment parameters during the spin turn for the hip osteoarthritis (HOA) group at baseline and the control group.

Parameter during spin turn		HOA Base-line	Control group	Mean difference	t-test/MWU
		Mean (SD)	Mean (SD)	(95%CI)	p (d)
Sagittal plane	Hip peak extension [°]	-20.08 (5.82)	-24.72 (5.25)	-4.64 (-8.09 -1.18)	0.010 (0.84)
	Hip peak flexion [°]	12.49 (5.82)	12.95 (5.55)	0.46 (-3.08 4.01)	0.793 (0.08)
	Hip sagittal RoM [°]	32.57 (6.49)	37.66 (5.25)	5.10 (1.42 8.78)	0.008 (0.86)
	Hip peak extension moment [Nm/kg]	2.45 (2.46)	2.50 (2.58)	0.27 (-0.39 0.93)	0.439 (0.02)
	Hip peak flexion moment [Nm/kg]	-2.89 (1.06)	-2.62 (1.06)	0.05 (-1.53 1.62)	0.411 (0.26)

Table 9.2: Results of discrete hip and knee joint angle and external joint moment parameters during the spin turn for the hip osteoarthritis (HOA) group at baseline and the control group.

Parameter during spin turn		HOA Base-line	Control group	Mean difference	t-test/MWU
		Mean (SD)	Mean (SD)	(95%CI)	p (d)
Sagittal plane	Knee peak extension [°]	11.90 (3.64)	10.06 (2.58)	-1.84 (-3.80 0.13)	0.066 (0.58)
	Knee peak flexion [°]	46.74 (5.15)	45.05 (5.24)	-1.68 (-4.92 1.56)	0.300 (0.32)
	Knee sagittal RoM [°]	34.84 (4.23)	34.99 (5.99)	0.15 (-3.08 3.39)	0.924 (0.03)
	Knee peak extension moment [Nm/kg]	-1.75 (1.80)	-1.80 (1.77)	-0.05 (-1.17 1.06)	0.901 (0.03)
	Knee peak flexion moment [Nm/kg]	3.06 (3.35)	3.45 (3.78)	0.38 (-1.84 2.61)	0.343 (0.11)
Frontal plane	Hip peak adduction [°]	-9.26 (3.12)	-11.25 (2.03)	-1.99 (-3.63 -0.35)	0.019 (0.75)
	Hip peak abduction [°]	1.95 (2.38)	1.71 (2.92)	-0.23 (-1.89 1.43)	0.777 (0.09)
	Hip frontal RoM [°]	11.21 (3.00)	12.97 (1.81)	1.76 (0.21 3.30)	0.027 (0.71)
	Hip peak adduction moment [Nm/kg]	-3.78 (3.63)	-4.28 (4.10)	-0.50 (-2.91 1.92)	0.550 (0.13)
	Hip peak abduction moment [Nm/kg]	1.09 (1.35)	1.33 (1.62)	0.25 (-0.68 1.18)	0.823 (0.17)
	Knee peak adduction [°]	-2.98 (3.65)	-4.63 (5.02)	-1.65 (-4.39 1.09)	0.230 (0.38)
	Knee peak abduction [°]	2.20 (4.56)	1.17 (4.33)	-1.02 (-3.79 1.75)	0.460 (0.23)
	Knee frontal RoM [°]	5.17 (2.02)	5.80 (2.73)	0.63 (-0.87 2.13)	0.601 (0.26)
	Knee peak adduction moment [Nm/kg]	0.69 (1.08)	1.12 (1.33)	0.43 (-0.33 1.18)	0.485 (0.35)
	Knee peak abduction moment [Nm/kg]	-2.63 (1.97)	-2.42 (2.08)	0.21 (-1.05 1.48)	0.501 (0.11)

Table 9.2: Results of discrete hip and knee joint angle and external joint moment parameters during the spin turn for the hip osteoarthritis (HOA) group at baseline and the control group.

Parameter during spin turn		HOA Base-line	Control group	Mean difference	t-test/MWU
		Mean (SD)	Mean (SD)	(95%CI)	p (d)
Transverse plane	Hip peak int. rotation [°]	-13.20 (12.19)	-19.62 (13.94)	-6.42 (-14.59 1.74)	0.120 (0.49)
	Hip peak ext. rotation [°]	15.82 (11.05)	15.41 (15.01)	-0.41 (-8.63 7.81)	0.920 (0.03)
	Hip transverse RoM [°]	29.02 (8.72)	35.03 (8.45)	6.01 (0.66 11.36)	0.029 (0.70)
	Hip peak internal rotation moment [Nm/kg]	-0.69 (0.83)	-0.71 (0.69)	-0.02 (-0.50 0.45)	0.765 (0.03)
	Hip peak external rotation moment [Nm/kg]	1.37 (1.06)	1.74 (1.22)	0.37 (-0.34 1.08)	0.184 (0.32)
	Knee peak int. rotation [°]	24.49 (11.28)	29.29 (13.55)	4.80 (-2.98 12.57)	0.220 (0.38)
	Knee peak ext. rotation [°]	49.65 (13.33)	46.66 (12.13)	2.98 (-4.97 10.93)	0.452 (0.23)
	Knee transverse RoM [°]	22.17 (6.70)	20.36 (3.81)	-1.81 (-5.21 1.59)	0.288 (0.33)
	Knee peak internal rotation moment [Nm/kg]	1.03 (0.60)	1.33 (0.71)	0.30 (-0.11 0.71)	0.083 (0.46)
	Knee peak external rotation moment [Nm/kg]	-0.71 (0.33)	-0.71 (0.48)	0.01 (-0.25 0.26)	0.971 (0.01)
Spatio-temporal	Stance phase duration [s]	0.78 (0.14)	0.74 (0.08)	-0.04 (-0.11 0.03)	0.251 (0.36)
	Forward CoM velocity at IC [m/s]	0.92 (0.19)	0.96 (0.12)	0.04 (-0.06 0.14)	0.396 (0.26)
	Forward CoM velocity at TO [m/s]	0.93 (0.12)	0.97 (0.11)	0.04 (-0.03 0.11)	0.303 (0.32)
	Foot progression angle at foot-flat	45.26 (14.39)	50.78 (14.23)	5.51 (-3.41 14.44)	0.219 (0.39)

Note. P-values as revealed by independent sample t-tests/Mann-Whitney-U tests (MWU) and effect sizes (|d|); significant results ($p < 0.05$) in bold. RoM = range of motion; CoM = centre of mass; IC = initial contact; TO = toe off.

9.4.2 Effects of hip bracing on hip and knee biomechanics

Results from the comparison of the HOA group under varying bracing conditions on spatiotemporal parameters and hip joint kinematics and dynamics during the step and spin turn are presented in Table 9.3 and Table 9.4, respectively. The corresponding results for the knee joint can be found in Table A.6 and Table A.7 in the Appendix.

For the step turn, forward-directed CoM velocity at IC was higher in both braced conditions compared to the baseline condition (short-term: MD = -0.07 m/s; $p = 0.013$; medium-term: MD = -0.13 m/s; $p = 0.004$) and also after medium-term bracing compared to short-term bracing (MD = -0.05 m/s; $p = 0.005$). The FPA at foot-flat was less inwardly rotated with short-term bracing compared to the baseline condition (MD = -7.96°; $p = 0.03$). Compared to baseline, peak hip flexion decreased with short-term bracing (MD = 2.66°; $p = 0.046$); peak hip internal rotation decreased with medium-term bracing (MD = -5.44°; $p = 0.024$); and hip transverse RoM decreased in both braced conditions (short-term: MD = 3.89°; $p < 0.001$; medium-term: MD = 4.00°; $p < 0.001$). No effects of bracing on knee biomechanics during the 90° step turn were found.

For the spin turn, forward-directed CoM velocity at IC was higher in both braced conditions compared to the baseline condition (short-term: MD = -0.10 m/s; $p = 0.006$; medium-term: MD = -0.14 m/s; $p = 0.001$) and also after medium-term bracing compared to short-term bracing (MD = -0.04 m/s; $p = 0.046$). The FPA at foot-flat was less outwardly rotated with short-term bracing compared to the baseline condition (MD = 7.21°; $p = 0.009$). For the hip joint, decreased transverse RoM was found in both braced conditions compared to baseline (short-term: MD = 2.94°; $p = 0.014$; medium-term: MD = 2.90°; $p = 0.025$). For the knee joint, compared to baseline, the peak knee adduction moment increased with short-term bracing (MD = -0.75 Nm/kg; $p = 0.015$), and peak internal rotation moment increased in both braced conditions (short-term: MD = -0.42 Nm/kg; $p = 0.021$; medium-term: MD = -0.43 Nm/kg; $p = 0.025$).

Table 9.3: Results of bracing effects on discrete spatio-temporal as well as kinematic and kinetic parameters of the hip joint during the step turn.

Parameter during step turn	Baseline		Short-term	Medium-term	ANOVA/ Friedman	B vs. S		B vs. M		S vs. M	
	Mean (SD)	Mean (SD)				MD (95%CI)	P	MD (95%CI)	P	MD (95%CI)	P
Sagittal plane	Hip peak extension [°]	-20.32 (5.73)	-21.30 (5.98)	-21.96 (5.19)	0.34 (0.05)	0.98 (-0.74 2.71)		1.64 (-0.79 4.07)		0.66 (-2.00 3.31)	
	Hip peak flexion [°]	8.18 (6.84)	5.52 (5.46)	5.95 (7.57)	0.03 (0.16)	2.66 (0.57 4.75)	0.046	2.23 (0.00 4.47)	0.101	-0.42 (-2.58 1.73)	0.687
	Hip sagittal RoM [°]	28.50 (8.01)	26.82 (7.50)	27.90 (9.02)	0.22 (0.07)	1.67 (-0.06 3.41)		0.59 (-1.77 2.96)		-1.08 (-2.87 0.71)	
	Hip peak extension moment [Nm/kg]	2.61 (2.74)	2.80 (3.12)	3.08 (3.27)	0.17 (0.11)	-0.11 (-0.69 0.46)		-0.17 (-0.65 0.30)		-0.06 (-0.50 0.38)	
Frontal plane	Hip peak flexion moment [Nm/kg]	-3.26 (2.53)	-2.97 (2.86)	-3.18 (3.21)	0.80 (0.01)	-0.29 (-1.22 0.64)		-0.08 (-0.95 0.78)		0.20 (-0.81 1.22)	
	Hip peak adduction [°]	5.47 (2.98)	6.12 (2.53)	6.03 (3.58)	0.33 (0.06)	-0.65 (-1.54 0.23)		-0.55 (-1.60 0.49)		0.10 (-0.86 1.05)	
	Hip peak abduction [°]	8.73 (2.39)	9.09 (2.92)	9.20 (3.90)	0.65 (0.02)	-0.36 (-1.35 0.63)		-0.47 (-1.68 0.74)		-0.11 (-1.21 0.99)	
	Hip frontal RoM [°]	-1.25 (1.47)	-1.63 (2.21)	-1.60 (2.25)	0.72 (0.09)	0.38 (0.01 0.76)		0.35 (-0.17 0.86)		-0.04 (-0.45 0.38)	
Frontal plane	Hip peak adduction moment [Nm/kg]	3.50 (3.01)	3.39 (3.18)	3.68 (3.25)	0.05 (0.06)	0.11 (-0.28 0.49)		-0.19 (-0.57 0.20)		-0.29 (-0.68 0.10)	
	Hip peak adduction moment [Nm/kg]	-3.26 (2.53)	-2.97 (2.86)	-3.18 (3.21)	0.80 (0.01)	-0.29 (-1.22 0.64)		-0.08 (-0.95 0.78)		0.20 (-0.81 1.22)	

Table 9.3: Results of bracing effects on discrete spatiotemporal as well as kinematic and kinetic parameters of the hip joint during the step turn.

Parameter during step turn		Baseline		Short-term	Medium-term	ANOVA/ Friedman		B vs. S		B vs. M		S vs. M	
		Mean (SD)	Mean (SD)	Mean (SD)	p (η ²)	MD (95%CI)	p	MD (95%CI)	p	MD (95%CI)	p		
Transverse plane	Hip peak int. rotation [°]	-13.82 (13.68)	-10.45 (10.84)	-8.38 (10.99)	0.02 (0.18)	-3.37 (-7.41 0.67)	0.195	-5.44 (-9.28 -1.59)	0.024	-2.07 (-5.66 1.52)	0.243		
	Hip peak ext. rotation [°]	6.97 (11.45)	6.45 (10.46)	8.41 (10.84)	0.51 (0.03)	0.53 (-2.96 4.01)		-1.44 (-5.11 2.23)		-1.97 (-5.77 1.84)			
	Hip transverse RoM [°]	20.79 (7.85)	16.89 (6.65)	16.79 (7.01)	<0.001 (0.51)	3.89 (2.31 5.47)	<0.001	4.00 (2.48 5.51)	<0.001	0.10 (-1.26 1.46)	0.878		
	Hip peak internal rotation moment [Nm/kg]	-1.81 (0.81)	-2.02 (0.98)	-1.96 (0.92)	0.17 (0.04)	0.21 (-0.19 0.60)		0.15 (-0.22 0.53)		-0.06 (-0.28 0.17)			
	Hip peak external rotation moment [Nm/kg]	0.58 (0.61)	0.62 (0.61)	0.66 (0.73)	0.95 (0.02)	-0.04 (-0.19 0.11)		-0.08 (-0.26 0.10)		-0.04 (-0.22 0.13)			
Spatio-temporal	Stance phase duration [s]	0.73 (0.10)	0.74 (0.12)	0.71 (0.12)	0.03 (0.06)	-0.01 (-0.03 0.02)	0.708	0.02 (-0.03 0.08)	0.222	0.03 (-0.02 0.09)	0.096		
	Forward CoM velocity at IC [m/s]	0.94 (0.14)	1.01 (0.19)	1.07 (0.21)	0.003 (0.31)	-0.07 (-0.14 -0.01)	0.013	-0.13 (-0.20 -0.05)	0.004	-0.05 (-0.09 -0.01)	0.005		

Table 9.3: Results of bracing effects on discrete spatio-temporal as well as kinematic and kinetic parameters of the hip joint during the step turn.

Parameter during step turn	Baseline		Short-term	Medium-term	ANOVA/ Friedman	B vs. S		B vs. M		S vs. M	
	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	P (η_p^2)	MD (95%CI)	P	MD (95%CI)	P	MD (95%CI)	P
Spatio-temporal	Forward CoM velocity at TO [m/s]	0.94 (0.16)	0.91 (0.14)	0.97 (0.14)	0.10 (0.12)	0.02 (-0.01 0.06)		-0.04 (-0.11 0.03)		-0.06 (-0.12 -0.01)	
	Foot progression angle at foot-flat [°]	-37.05 (15.88)	-29.09 (15.22)	-31.39 (16.16)	0.04 (0.15)	-8.96 (-15.2 -0.70)	0.030	-5.66 (-12.9 1.61)	0.119	2.29 (-2.67 1.4)	0.320

Note: P-values as revealed by one-way repeated measures ANOVAs / Friedman tests and Holm-Bonferroni corrected pairwise comparisons and effect sizes (η_p^2); significant results ($p < 0.05$) in bold. MD = Mean difference; 95%CI = 95 % confidence intervals; B = baseline; S = short-term; M = medium-term; RoM = Range of Motion; CoM = centre of mass; IC = Initial contact; TO = toe off.

Table 9.4: Results of bracing effects on discrete spatiotemporal as well as kinematic and kinetic parameters of the hip joint during the spin turn.

Parameter during spin turn		Baseline		Short-term	Medium-term	ANOVA/ Friedman	B vs. S		B vs. M		S vs. M	
		Mean (SD)	Mean (SD)	Mean (SD)	p (η_p^2)		MD (95%CI)	p	MD (95%CI)	p	MD (95%CI)	p
Sagittal plane	Hip peak extension [°]	-20.08 (5.82)	-21.87 (5.48)	-21.54 (5.15)	0.15 (0.09)	1.79 (0.05,3.53)		1.46 (-0.70,3.61)		-0.33 (-2.36,1.70)		
	Hip peak flexion [°]	12.49 (5.82)	11.00 (6.13)	11.32 (6.63)	0.28 (0.06)	1.49 (-0.32,3.30)		1.17 (-0.98,3.32)		-0.32 (-2.38,1.74)		
	Hip sagittal RoM [°]	32.57 (6.49)	32.87 (7.17)	32.86 (6.47)	0.87 (0.01)	-0.30 (-1.86,1.26)		-0.29 (-1.46,0.88)		0.01 (-1.31,1.34)		
	Hip peak extension moment [Nm/kg]	2.45 (2.46)	2.67 (2.73)	2.33 (2.49)	0.83 (0.06)	-0.26 (-0.71,0.20)		0.05 (-0.70,0.81)		0.31 (-0.28,0.90)		
Frontal plane	Hip peak flexion moment [Nm/kg]	-2.89 (1.06)	-2.63 (0.76)	-2.94 (1.54)	0.54 (0.03)	-0.23 (-0.67,0.22)		0.12 (-0.30,0.54)		0.34 (-0.16,0.85)		
	Hip peak adduction [°]	-9.26 (3.12)	-8.26 (3.20)	-8.94 (3.14)	0.04 (0.15)	-1.00 (-1.81,-0.19)	0.053	-0.33 (-1.11,0.45)	0.388	0.67 (-0.12,1.47)	0.186	
	Hip peak abduction [°]	1.95 (2.38)	2.91 (2.84)	2.72 (3.24)	0.16 (0.09)	-0.96 (-2.01,0.09)		-0.78 (-2.04,0.48)		0.18 (-0.74,1.10)		
	Hip frontal RoM [°]	11.21 (3.00)	11.17 (3.28)	11.66 (3.97)	0.95 (0.02)	0.04 (-1.12,1.21)		-0.45 (-1.86,0.96)		-0.49 (-1.31,0.34)		
	Hip peak abduction moment [Nm/kg]	-3.78 (3.63)	-3.72 (3.88)	-3.95 (4.01)	0.13 (0.03)	-0.06 (-0.50,0.38)		0.16 (-0.33,0.65)		0.22 (-0.08,0.53)		
Frontal plane	Hip peak adduction moment [Nm/kg]	1.09 (1.35)	1.56 (1.82)	1.52 (1.89)	0.40 (0.10)	-0.47 (-1.04,0.09)		-0.44 (-1.11,0.24)		0.03 (-0.17,0.24)		

Table 9.4: Results of bracing effects on discrete spatio-temporal as well as kinematic and kinetic parameters of the hip joint during the spin turn.

Parameter during spin turn		Baseline	Short-term	Medium-term	ANOVA/ Friedman	B vs. S		B vs. M		S vs. M	
		Mean (SD)	Mean (SD)	Mean (SD)	p (η_p^2)	MD (95%CI)	p	MD (95%CI)	p	MD (95%CI)	p
Transverse plane	Hip peak int. rotation [°]	-13.20 (12.19)	-11.92 (11.26)	-9.75 (12.06)	0.18 (0.08)	-1.28 (-5.38 2.83)		-3.44 (-7.31 0.42)		-2.17 (-5.66 1.33)	
	Hip peak ext. rotation [°]	15.82 (11.05)	14.16 (8.86)	16.37 (10.67)	0.37 (0.05)	1.66 (-1.37 4.70)		-0.55 (-4.01 2.91)		-2.21 (-5.75 1.33)	
	Hip transverse RoM [°]	29.02 (8.72)	26.08 (8.87)	26.12 (10.38)	0.01 (0.23)	2.94 (1.02 4.87)	0.014	2.90 (0.70 5.10)	0.025	-0.05 (-2.08 1.98)	0.962
	Hip peak internal rotation moment [Nm/kg]	-0.69 (0.83)	-0.69 (0.75)	-0.69 (0.75)	0.54 (0.00)	0.01 (-0.18 0.19)		0.01 (-0.20 0.22)		0.00 (-0.12 0.13)	
Spatio-temporal	Hip peak external rotation moment [Nm/kg]	1.37 (1.06)	1.67 (1.27)	1.76 (1.33)	0.08 (0.24)	-0.30 (-0.55 -0.06)		-0.40 (-0.66 -0.13)		-0.09 (-0.31 0.12)	
	Stance phase duration [s]	0.78 (0.14)	0.82 (0.16)	0.77 (0.16)	0.08 (0.08)	-0.04 (-0.09 0.00)		0.01 (-0.06 0.09)		0.05 (-0.01 0.12)	
	Forward CoM velocity at IC [m/s]	0.92 (0.19)	1.02 (0.20)	1.05 (0.22)	<0.001 (0.36)	-0.10 (-0.16 -0.03)	0.006	-0.14 (-0.22 -0.06)	0.001	-0.04 (-0.07 0.00)	0.046

Table 9.4: Results of bracing effects on discrete spatiotemporal as well as kinematic and kinetic parameters of the hip joint during the spin turn.

Parameter during spin turn		Baseline	Short-term	Medium-term	ANOVA/ Friedman	B vs. S		B vs. M		S vs. M	
		Mean (SD)	Mean (SD)	Mean (SD)	p (η_p^2)	MD (95%CI)	p	MD (95%CI)	p	MD (95%CI)	p
Spatio-temporal	Forward CoM velocity at TO (m/s)	0.93 (0.12)	0.89 (0.13)	0.91 (0.15)	0.32 (0.05)	0.04 (0.00,0.08)		0.02 (-0.05, 0.08)		-0.02 (-0.07, 0.02)	
	Foot progression angle at foot-flat [°]	45.26 (14.39)	38.05 (12.15)	38.47 (15.22)	0.04 (0.18)	7.21 (2.75,11.68)	0.009	6.80 (-0.77, 14.37)	0.152	-0.42 (-4.86,4.03)	0.847

Note. P-values as revealed by one-way repeated measures ANOVAs / Friedman tests and Holm-Bonferroni corrected pairwise comparisons and effect sizes (η_p^2); significant results ($p < 0.05$) in bold. MD = Mean difference; 95%CI = 95 % confidence intervals; B = baseline; S = short-term; M = medium-term; RoM = Range of Motion; CoM = centre of mass; IC = initial contact; TO = toe off.

9.5 Discussion

9.5.1 Effects of hip osteoarthritis on hip and knee biomechanics

The first aim of this study was to quantify the effect of mild-to-moderate HOA on hip and knee biomechanics during 90° turns while walking. As hypothesised, HOA participants demonstrated reduced peak hip extension and reduced sagittal and transverse hip RoM in both turning conditions, as well as reduced peak hip adduction, resulting in a decreased frontal hip RoM during the spin turn. Additionally, increased knee flexion was found during the step turn in the HOA group compared to the CON group. Subsequent analyses showed that larger transverse hip RoM during the step turn and larger peak hip extension during the spin turn were moderately correlated with better hip function based on the HHS in persons with HOA.

The limitations observed in sagittal and transverse hip kinematics during turning are in line with those occurring during straight walking (Benedetti et al., 2012; Constantinou et al., 2017; Eitzen et al., 2012; Kumar et al., 2015; Leigh et al., 2016; Rutherford et al., 2015a; Steingrebe et al., 2022, 2023) in persons with mild-to-moderate HOA. Thereby, the limitations in sagittal and transverse hip RoM mainly stem from reduced hip extension and internal rotation angles. Spin and step turning has been found to require substantially larger internal and external rotation at the knee joint (Zürcher et al., 2008) as well as larger external rotation angles at the hip joint (M. M. Taylor et al., 2005) than straight walking. Compared to data from straight walking in the same cohort (Steingrebe et al., 2022), we observed larger peak hip internal rotation angles during both turning conditions, as well as larger hip external rotation during the spin turn. Thus, limitations in transverse plane hip mobility due to HOA are more likely to impose difficulties during 90° turning than during straight walking, which was also shown by the correlation between hip function and transverse hip RoM during the step turn. Likewise, reduced hip adduction during the spin turn is in line with data from 45° turns in persons with severe HOA (Tateuchi et al., 2014); but was not present during straight walking in persons with mild-to-moderate HOA (Constantinou et al., 2017). In contrast to our expectations, we did not find reduced peak hip abduction during the step turn. However, in the study of Tateuchi et al. (Tateuchi et al., 2014), participants suffered from end-stage HOA, so alterations in hip abduction angle during the step turn might depend on HOA severity. Therefore, including 90° spin and step turns in clinical gait analysis might be helpful to

identify limitations in frontal and transverse plane hip mobility that might not be apparent in straight walking, and the hip transverse RoM could potentially differentiate between various degrees of HOA.

The increase in knee flexion during the step turn contradicts the results from 45° spin turns, as well as straight walking (Tateuchi et al., 2014) where a reduction of peak knee flexion was described. Increasing knee flexion might allow people to lower the CoM and thereby increase CoM stability during turning, at the cost of higher muscular demands for the quadriceps muscle. Furthermore, flexion of the knee joint allows greater knee external rotation due to the relaxation of the joint capsule and collateral ligaments (Zarins et al., 1983). Thus, flexion of the knee joint might be a mechanism to compensate for a lack of hip external rotation, which is needed in more pronounced turning of 90° compared to 45° turns. Furthermore, in contrast to Tateuchi et al. (2014), no differences in joint dynamics were found between the HOA and CON groups. However, as stated previously, the population in their study consisted of patients with severe HOA, which may have affected gait dynamics, as these have been found to depend on HOA severity (Diamond et al., 2018). Thus, alterations in joint dynamics might not be present in people with mild-to-moderate HOA.

9.5.2 Effects of hip bracing on hip and knee biomechanics

The second aim of this study was to quantify the short- and medium-term effects of hip bracing on hip and knee biomechanics during 90° turns while walking. The main findings are that bracing led to an increase in initial movement velocity as well as a short-term adjustment of foot placement strategy in both turning conditions. In line with our hypothesis, transverse hip RoM was reduced in both braced conditions and during both turning movements. In contrast to our expectations, no differences in frontal hip biomechanics were found. However, hip bracing led to increases in knee adduction and internal rotation moments during the spin turn.

Increasing movement velocity at turn initiation could be interpreted as an increase in movement confidence, possibly explained by decreased pain during movement. Yet, as randomisation of bracing conditions was not possible, it might also result from habituation with the movement task.

E. Sato et al. (2012) observed increased TUG velocity with unilateral bracing, but only if turning was conducted with the unbraced leg inside. This suggests that the increase in velocity stemmed from alterations in the turning movement and not during rising, walking or sitting

down. It was speculated that the beneficial effect of bracing is more pronounced when the hip joint is abducted and externally rotated. In contrast to these findings, an increase in initial CoM velocity was present in both turning conditions in our study. However, the TUG requires a 180° change in direction, which is likely executed with more than one step, and thus not directly comparable to the 90° turns conducted in our study. Additionally, it has to be noted that the increase in movement initiation velocity did not reduce ST duration. Higher movement velocities are linked to higher GRF and subsequently higher joint moments (Bejek et al., 2006; Zeni & Higginson, 2009). However, despite an increased CoM velocity at turn initiation, no changes in hip moments were found in either turning condition. During the step turn, greater knee adduction and internal rotation moments were observed, which might indicate additional loading to be compensated at the knee rather than the hip joint.

The passive resistance of the brace might explain the decrease in peak hip flexion during the step turn and the reduction of the inward and outward rotation of the foot during the step and spin turns, respectively. Yet, all of these changes diminish after medium-term brace application due to habituation.

Data from the CON group show that larger hip internal rotation is needed during the step compared to the spin turn. However, similar peak hip internal rotation angles are found during both turning conditions in the HOA group. Thus, both turning conditions might require the entire available functional transverse hip RoM of the HOA participants. After medium-term bracing, decreased hip internal rotation during the step turn was observed. Thus, bracing might prevent RoM endpoints from being reached, which might be associated with hip pain (Reiman et al., 2020; St-Pierre et al., 2022). Similar results have previously been found by Nérot & Nicholls (2017) during walking. However, the brace in their study aimed to alter hip biomechanics by mechanically induced abduction and outward rotation of the femur, which is not the case for the brace applied in the current study, which aims to manipulate soft tissue around the hip joint and pelvis.

9.5.3 Limitations

We acknowledge several limitations in this study. First, during our study, the starting position of the participants in the lab was manipulated to provoke either a spin or a step turn on the force plate. Thus, no data exists on the self-selected turning strategy. In our subjective perception during data capture, participants tended to alter step length to complete the turn in a step-turn manner. Yet, this is speculative at best. Glaister et al. (2007) stated that turning in

most activities of daily living happens in a step-turn manner, and likewise, data from Dixon et al. (2019) suggest that pre-planned turns on flat surfaces are mainly conducted in this manner. In contrast, Akram et al. (2010) reported a high number of spin turns in healthy older adults. As no data exist on the preferred turning strategies in people with HOA, the turning preferences in this population are unknown. To estimate the impact of brace-induced modifications in step and spin turn biomechanics, the relevance of these movements for the daily loading profile of persons with HOA has to be clarified in the future. However, from a diagnostic perspective, spin turning in our study required more pronounced hip adduction and external rotation as well as larger transverse RoM of the hip and knee joints and can, therefore, serve as a valuable addition to straight walking in clinical gait analysis.

Secondly, interpretation of transverse plane hip and knee biomechanics should always happen with caution because multi-body modelling is prone to errors in this movement plane. Observed knee transverse angles in the present study are shifted towards external rotation compared to the literature. This offset is caused by differences in the definition of the tibia neutral position. Consequently, when comparing our data to other studies, individual model definitions should be considered. However, transverse plane joint movements are of high significance for executing rotational movements such as turning while walking and should therefore not be disregarded. Furthermore, the error is systematic and should not impact group comparison. The observed group differences in the transverse plane hip angle ranged between 6 and 10° and thus should not be dismissed.

9.6 Conclusion

While limitations in hip mobility have already been investigated during 45° turns in persons with end-stage HOA (Tateuchi et al., 2014), the biomechanics during potentially more relevant (Sedgman et al., 1994) 90° turns, in persons with mild-to-moderate HOA, were unknown. In the present study, this population demonstrated limited hip mobility in all three movement planes during 90° spin turns. While partly not apparent during straight walking, mobility limitations of hip internal and external rotation, as well as hip adduction, become especially apparent during this movement task. Therefore, the inclusion of 90° spin turns in the portfolio of clinical movement analysis might allow a more comprehensive understanding of movement limitations at an earlier stage of HOA, and thus improve planning and evaluation of conservative treatment options. Hip bracing, as one conservative treatment option, led to higher movement velocity at turn initiation and reduced transverse hip RoM during 90° turns. While no increase in hip joint moments was observed, increases in knee joint load-

ing warrant further investigation to avoid overloading of adjacent joints. Additionally, future studies should assess the rate of occurrence of 90° turns in the everyday life of persons with HOA as well as their preferred coping strategy (spin vs. step turn) to further clarify the ecological validity of this movement task.

10 Manuscript VII - Effect of hip bracing on stair walking biomechanics and pain

Slightly modified version submitted to *BMC Musculoskeletal Disorders* and currently under review for publication as:

Steingrebe, H., Ehmann, H., Sell, S., & Stein, T. Effect of hip bracing on stair walking biomechanics and pain in patients with mild-to-moderate hip osteoarthritis: an intervention study.

10.1 Abstract

Bracing is a conservative treatment method for HOA and has shown favourable effects on pain and functional capacity. However, biomechanical analyses of brace effects remain sparse and are limited to level walking. Stair walking is more demanding than level walking in terms of movement coordination and joint loads. This study, therefore, aimed to investigate the effect of hip bracing on pain perception and biomechanics of the hip, pelvis, and trunk during stair walking in individuals with HOA.

Hip, pelvis, and trunk biomechanics and pain during stair ascent and descent were assessed before and after one week of hip bracing in 20 individuals with unilateral mild-to-moderate HOA. Differences between the bracing conditions were analysed with dependent t-tests, and Pearson's correlations were used to analyse the correlation between brace-induced alterations in pain score and biomechanical parameters.

Bracing increased movement velocity and reduced stair walking pain by 28 %. Furthermore, increased hip extension and reduced hip flexion were found with bracing. Bracing led to a more upright pelvis position, but did not affect trunk motion. During stair ascent, frontal pelvis motion increased, while peak hip adduction and internal rotation decreased with bracing. During stair descent, increased hip extension and external rotation moments were found with bracing, while the pelvis and hip transverse RoM were reduced. Decreased pelvis rise on

the ipsilateral side during stair ascent and increased hip transverse RoM during stair descent were moderately correlated with a decrease in pain.

Bracing can reduce hip pain during stair walking and mitigate some of the effects of HOA on stair walking biomechanics. Bracing led to a more upright pelvis position, combined with increased hip extension and decreased hip flexion angles, and could potentially slow HOA progression. In contrast, hip bracing exacerbated the reduction in hip adduction observed in individuals with HOA and induced additional changes, such as increased pelvis oblique motion.

10.2 Introduction

Hip osteoarthritis is a chronic, progressive disease whose prevalence increases with age (Dagenais et al., 2009). Despite its widespread occurrence, recommendations for HOA management are often extrapolated from KOA research (Bannuru et al., 2019; N. Murphy et al., 2016). Furthermore, the majority of HOA research has concentrated on addressing end-stage HOA. By shifting the management focus to the earlier stages of the disease, it may be possible to slow disease progression, enhance patient quality of life, and ultimately reduce the number of hip replacement surgeries (N. Murphy et al., 2016). Conservative management of HOA aims to reduce joint pain and stiffness, maintain or improve joint mobility and stability, enhance activities and participation, and improve the quality of life (Bannuru et al., 2019; Fernandes et al., 2013). Thereby, it has been shown that individuals with HOA have a strong desire to self-manage their condition (Claassen et al., 2019).

One option for conservative HOA management that can easily be regulated by the patient is hip bracing. Hip bracing for HOA has shown favourable effects on pain and functional capacity (Nérot & Nicholls, 2014; E. E. Sato et al., 2012; Steingrebe et al., 2022). However, biomechanical analyses of bracing effects in HOA are sparse and limited to level walking movements (Nérot & Nicholls, 2017; Steingrebe et al., 2022). It has been shown that a brace aiming to decrease load at the hip joint reduced the peak hip adduction and internal rotation angles as well as the peak hip abduction moment (Nérot & Nicholls, 2017). Using an elastic brace resulted in increased gait speed, step length, and peak hip extension moments (Steingrebe et al., 2022).

Although walking on level ground is one of the most fundamental daily movements, walking up and down stairs is more demanding due to the greater sagittal and frontal joint angles of the lower limbs, greater moments at the hip and knee joints (Andriacchi et al., 1980; Nadeau et al., 2003; Riener et al., 2002), and greater hip joint contact forces (Bergmann et al., 2001). This high demand leads to lower performance during stair walking tests of individuals with various stages of HOA severity compared to healthy controls (M. M. Arokoski et al., 2004; Burgess et al., 2022; Judd et al., 2014). Furthermore, stair walking ability decreased with HOA progression (M. M. Arokoski et al., 2004; Stasi et al., 2023).

Stair walking in individuals with various degrees of HOA differs from healthy controls in terms of lower movement velocity, increased trunk flexion towards the affected limb and lower hip RoM in all planes, caused by lower hip flexion, extension, abduction, adduction and external rotation angles (Hall et al., 2017; Meyer et al., 2016; Popovic et al., 2021; Steingrebe et al., 2025). Additionally, in one study, increased hip internal rotation angles were found (Popovic et al., 2021). Furthermore, hip extension, adduction and internal and external rotation moments (Hall et al., 2017; Meyer et al., 2016), as well as peak hip contact forces (van Rossom et al., 2023), are lower in individuals with HOA.

Thus, stair walking motion is impaired at all levels of HOA. However, to date, it remains unclear whether hip bracing can influence stair walking biomechanics and enhance stair walking ability. Therefore, the primary objective of the present study was to evaluate the impact of bracing on hip pain and biomechanics of the hip, pelvis, and trunk during stair walking in individuals with mild-to-moderate HOA. We expected bracing to reduce hip pain during stair walking and reverse the impact of mild-to-moderate HOA by increasing movement velocity, frontal hip mobility and hip internal rotation moments, while decreasing hip internal rotation angles and ipsilateral trunk lean. The second aim of this study was to explore which brace-induced biomechanical changes were associated with changes in pain perception during stair walking.

10.3 Materials and Methods

Data for this intervention study were collected between April 2019 and March 2021 in the BioMotion Center of the IfSS at the KIT.

10.3.1 Participants

The cohort for this study consisted of 21 individuals (10 females; age: 64.0 ± 9.6 years; BMI: 24.2 ± 2.9) with symptomatic mild-to-moderate unilateral HOA. Participants had radiographically confirmed HOA of K-L grade 2-4 and experienced hip pain during daily activities over the past three months. Due to the weak association between radiographic HOA features and functional impairments (Pereira et al., 2016), mild-to-moderate symptoms were defined functionally by a Harris Hip Score ranging from 65 to 95 (74.6 ± 11.8). None of the participants in our study reported experiencing extreme pain while walking up or down stairs, as measured by the HOOS (A. K. Nilsson et al., 2003). Additionally, all participants were able to ascend and descend stairs in a step-over manner without using the handrail.

The contralateral limb needed to be pain-free, possess a K-L grade ≤ 2 , and demonstrate an unrestricted passive joint RoM (sagittal RoM $\geq 90^\circ$, transverse RoM $\geq 15^\circ$, peak abduction $\geq 20^\circ$, flexing contracture $\leq 10^\circ$). Exclusion criteria included secondary HOA resulting from trauma, neuromuscular disorders or neurological complaints, a BMI of $\geq 35 \text{ kg/m}^2$, or other orthopaedic injuries of the lower limbs and back (Table 6.1).

The ethics committee at the KIT approved the study procedure. All participants provided written informed consent prior to participating in the study.

10.3.2 Hip brace

The hip brace used in this study (Coxatrain®, Bauerfeind AG; Figure 8.2) comprises three components: an elastic pelvis belt, an elastic thigh belt, and a rigid connecting piece featuring a hinge joint that permits one degree of freedom (flexion/extension). Within the pelvis belt, viscoelastic pads are situated around the SIJ and the GMed regions. The rigid connecting piece includes a fixed pad at the greater trochanter and a movable pad positioned above it. The movable pad shifts vertically as the hinge joint flexes and extends. The purpose of the brace is to stabilise the pelvis girdle and SIJ, alleviate pain through trigger point massage of the SIJ and GMed regions, and restore functional hip mobility through continuous friction of GMed.

10.3.3 Testing protocol

Participants visited the lab three times: on the first occasion, baseline gait analysis was conducted without bracing. Subsequently, participants recorded their pain perception during stair walking on a 10 cm visual analogue scale (VAS; anchored: “no pain” to “worst imaginable pain”) once a day for one week (baseline period). During the second visit, participants were individually fitted with a hip brace. Thereafter, participants wore the brace for one week (prescribed wear time > 4 hours per day) during everyday activities, and simultaneously recorded their pain perception during stair walking (braced period). During the third visit, gait analysis was repeated while wearing the brace (braced). In each analysis, participants performed five valid trials (no handrail use, no tripping) of stair ascent and descent at a self-selected speed in a step-over manner. Stair ascent was initiated one or two steps in front of the staircase at ground level, while stair descent began without approach steps from the top platform of the staircase. VAS pain scores were averaged across the seven days of each period.

10.3.4 Data acquisition, data processing and biomechanical modelling

Hip, pelvis, and trunk biomechanics were assessed using a 16-camera infrared motion capture system (200 Hz; Vicon Motion Systems, Oxford Metrics Group, Oxford, UK) and a custom-built instrumented staircase with six steps (riser height = 17 cm, tread width = 28 cm, inclination angle = 31.26° ; Figure 10.1). Steps one to three and six were instrumented with 1D strain gauges for segmentation purposes (1000 Hz, Wii Balance Board, Nintendo, Kyoto, Japan), while steps four and five were equipped with 3D strain gauges (1000 Hz; K3D120 $\pm$ 1kN; ME-Meßsysteme GmbH, Henningsdorf, Germany). Participants were equipped with a full-body marker set consisting of 42 retroreflective markers, and 65 anthropometric measures were taken for individual scaling of the biomechanical model (Hermsdorf et al., 2019). In the braced condition, the ASIS were covered by the brace, necessitating the application of markers on the brace. The markers on the PSIS were affixed through the mesh fabric of the brace.

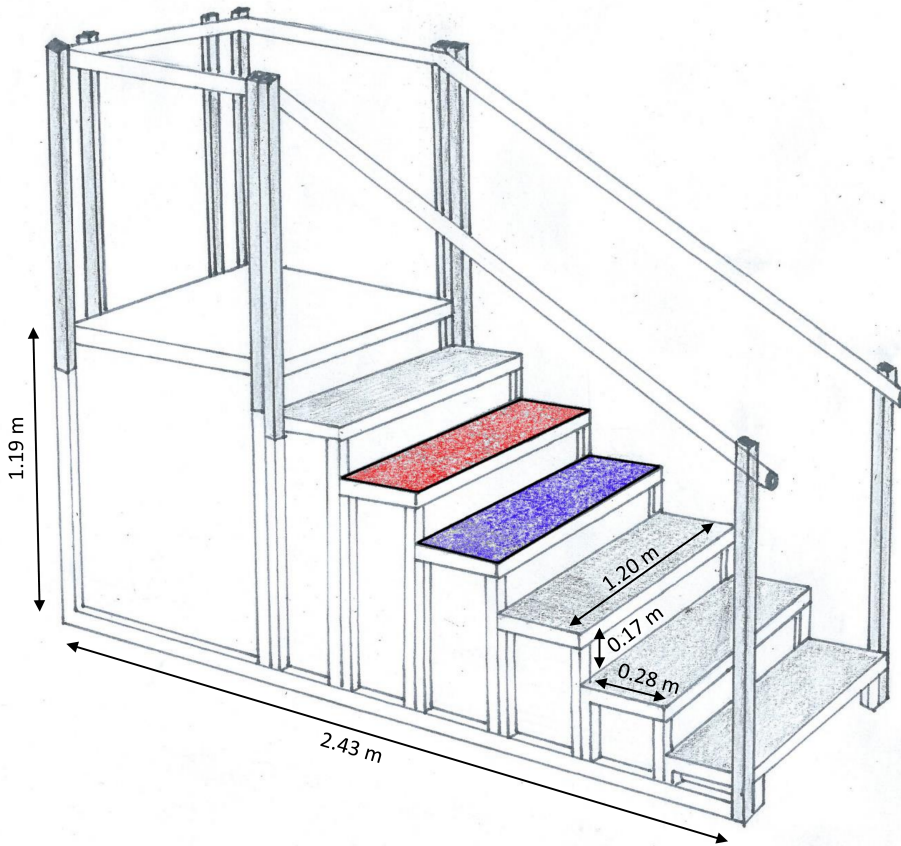


Figure 10.1: Sketch of the instrumented staircase. Three-dimensional force plates in steps four (blue) and five (red).

Note. Riser height = 17 cm, tread = 28 cm, inclination angle = 31.26° .

Postprocessing of the data was conducted in Vicon Nexus (version 2.14.0) and MATLAB (version R2022a; The MathWorks Inc., Natick, MA, USA). Kinematic and GRF data were filtered using a 4th-order Butterworth low-pass filter with a cut-off frequency of 10 Hz (Kristianslund et al., 2012). 3D hip joint angles and external joint moments, as well as pelvis orientation, were calculated using an inverse kinematics and dynamics approach with the multi-body model ALASKA Dynamicus (Hermsdorf et al., 2019), including the HJC definition proposed by Harrington (Harrington et al., 2007). Hip joint moments were normalised to body mass. Additionally, sagittal and frontal trunk angles were determined as the angle formed between a

vector extending from the midpoint of both ASIS markers to the clavicle marker and the vertical axis. The evaluated GC started with IC on the 4th step and ended with IC on the 6th step for stair ascent, or started with IC on the 5th step and ended with IC on the 3rd step for descent. IC was detected by a vertical GRF > 15 N. The dependent variables were peak hip joint angles, hip RoM and peak hip joint moments, peak pelvis orientation angles and RoM, peak sagittal and frontal trunk angles, mean vertical and horizontal CoM velocity, GC duration, and ST duration.

Brace wear time was assessed using temperature sensors (Orthotimer®, Rollerwerk Medical Engineering and Consulting, Balingen, Germany) on the inside of the pelvis belt. Sensor data were cross-checked with a wear diary self-recorded by the participants.

10.3.5 Statistical analysis

The required sample size was estimated through *a priori* power analysis using G*Power (version 3.1.9.3; (Faul et al., 2007)), based on data from Nérot & Nicholls (2017). They observed a bracing effect of 0.92 and 1.17 during walking for peak hip adduction and internal rotation, respectively. Consequently, with $\alpha = 0.05$ and power = 0.95, the minimum sample size necessary was between 12 and 18 subjects.

Statistical analyses were performed using R (version 4.2.2; (R Core Team, 2023)). The normal distribution of residuals was assessed with the Shapiro-Wilk test. Effects were analysed using t-tests for dependent samples or the WCX when the normal distribution of residuals could not be assumed. The level of significance was set *a priori* to $\alpha < 0.05$. Cohen's d was used to estimate the effect size and was interpreted as $|d| \geq 0.2$ a small effect, $|d| \geq 0.5$ a moderate effect and $|d| \geq 0.8$ a large effect (Cohen, 1988).

To evaluate the relationship between changes in pain perception and biomechanical variables, Pearson's correlations were calculated between the change in pain score and the change in biomechanical variables significantly affected by brace application. Correlation coefficients were interpreted as $|r| \geq 0.1$ a small effect, $|r| \geq 0.3$ a moderate effect and $|r| \geq 0.5$ a large effect (Cohen, 1988).

Due to a technical error, the GRF data for one of the 21 participants were not recorded during the third session. Consequently, all analyses were conducted using the data from the remaining 20 participants.

10.4 Results

Table 10.1 and Table 10.2 present descriptive data for all biomechanical parameters, along with the results of the analyses for stair ascent and descent, respectively. Time curves of the hip and pelvis angles in the sagittal and frontal planes are shown in Figure 10.2 and Figure 10.3. Time curves for the hip angle in the transverse plane and 3D hip moments are included in the Appendix (Figure A.32 to Figure A.35).

Table 10.1: Spatiotemporal, hip, pelvis, and trunk parameters during stair ascent at baseline and after one week of brace application (braced).

Parameter during stair ascent	Baseline	Braced	Mean difference	t-test/WCX
	Mean (SD)	Mean (SD)	(95%CI)	p (d)
Peak hip extension [°]	-9.50 (4.57)	-12.77 (4.97)	3.27 (1.24 5.30)	0.003 (0.68)
Peak hip flexion [°]	48.61 (6.80)	43.79 (7.45)	4.82 (3.47 6.16)	<0.001 (0.66)
Hip sagittal range of motion [°]	58.11 (5.55)	56.56 (6.02)	1.55 (-0.02 3.12)	0.053 (0.26)
Peak hip extension moment [Nm/kg]	0.53 (0.38)	0.49 (0.38)	0.04 (-0.10 0.18)	0.559 (0.10)
Peak hip flexion moment [Nm/kg]	-1.13 (0.41)	-1.07 (0.49)	-0.06 (-0.22 0.09)	0.143 (0.14)
Peak backward trunk lean [°]	-3.07 (14.27)	-1.89 (12.86)	-1.18 (-3.00 0.65)	0.194 (0.08)
Peak forward trunk lean [°]	0.79 (14.50)	2.14 (12.93)	-1.35 (-3.15 0.45)	0.133 (0.09)
Peak pelvis posterior tilt [°]	9.82 (3.72)	6.86 (3.57)	2.96 (1.61 4.30)	<0.001 (0.81)
Peak pelvis anterior tilt [°]	14.71 (3.38)	12.76 (2.83)	1.95 (0.61 3.29)	0.007 (0.62)
Pelvis sagittal range of motion [°]	4.90 (1.71)	5.90 (2.02)	-1.01 (-1.84 -0.18)	0.020 (0.53)
Peak hip adduction [°]	-9.03 (4.02)	-7.43 (3.08)	-1.60 (-2.70 -0.50)	0.007 (0.42)
Peak hip abduction [°]	3.38 (2.83)	4.41 (3.00)	-1.03 (-2.18 0.12)	0.076 (0.35)
Hip frontal range of motion [°]	12.41 (3.54)	11.84 (3.15)	0.57 (-0.54 1.67)	0.294 (0.17)
Peak hip adduction moment [Nm/kg]	-0.48 (0.21)	-0.53 (0.22)	0.04 (-0.06 0.15)	0.729 (0.20)
Peak hip abduction moment [Nm/kg]	0.19 (0.14)	0.18 (0.13)	0.01 (-0.03 0.05)	0.927 (0.09)
Peak contralateral trunk lean [°]	-2.59 (1.07)	-2.59 (1.49)	0.00 (-0.75 0.75)	0.997 (0.00)
Peak ipsilateral trunk lean [°]	2.26 (2.31)	2.77 (1.93)	-0.51 (-1.38 0.36)	0.236 (0.24)
Peak ipsilateral pelvis drop [°]	-5.12 (2.03)	-6.43 (1.83)	1.31 (0.56 2.06)	0.002 (0.68)

Table 10.1: Spatiotemporal, hip, pelvis, and trunk parameters during stair ascent at baseline and after one week of brace application (braced).

Parameter during stair ascent	Baseline	Braced	Mean difference	t-test/WCX
	Mean (SD)	Mean (SD)	(95%CI)	p (d)
Peak ipsilateral pelvis rise [°]	3.91 (1.75)	4.70 (2.27)	-0.79 (-1.54 -0.04)	0.040 (0.37)
Pelvis frontal range of motion [°]	9.03 (2.70)	11.13 (3.10)	-2.10 (-3.17 -1.03)	0.001 (0.72)
Peak hip int. rotation [°]	-4.00 (10.76)	-0.33 (10.07)	-3.67 (-7.13 -0.20)	0.039 (0.35)
Peak hip ext. rotation [°]	6.40 (11.07)	9.44 (10.61)	-3.03 (-7.17 1.10)	0.330 (0.27)
Hip transverse range of motion [°]	10.40 (1.87)	9.77 (3.67)	0.63 (-1.01 2.28)	0.432 (0.21)
Peak hip internal rotation moment [Nm/kg]	-0.13 (0.07)	-0.14 (0.06)	0.01 (-0.02 0.04)	0.447 (0.16)
Peak hip external rotation moment [Nm/kg]	0.17 (0.06)	0.17 (0.07)	0.00 (-0.02 0.03)	0.875 (0.03)
Pelvis transverse range of motion [°]	6.78 (2.10)	6.95 (2.76)	-0.18 (-1.51 1.16)	0.786 (0.07)
Stance phase duration [s]	0.79 (0.14)	0.75 (0.12)	0.05 (0.02 0.08)	0.005 (0.34)
Gait cycle duration [s]	1.23 (0.19)	1.17 (0.17)	0.06 (0.02 0.11)	0.010 (0.34)
Mean horizontal CoM velocity [m/s]	0.47 (0.07)	0.49 (0.08)	-0.02 (-0.04 0.00)	0.021 (0.27)
Mean vertical CoM velocity [m/s]	0.30 (0.04)	0.31 (0.05)	-0.01 (-0.02 0.00)	0.021 (0.25)

Note. External hip moments. P-values as revealed by dependent sample t-tests/Wilcoxon tests (WCX) and effect sizes (|d|); significant results ($p < 0.05$) in bold. 95%CI = 95 % confidence interval.

Table 10.2: Spatiotemporal, hip, pelvis, and trunk parameters during stair descent at baseline and after one week of brace application (braced).

Parameter during stair descent	Baseline	Braced	Mean difference	t-test/WCX
	Mean (SD)	Mean (SD)	(95%CI)	p (d)
Peak hip extension [°]	-6.13 (5.74)	-8.73 (7.12)	2.60 (0.71 4.49)	0.009 (0.38)
Peak hip flexion [°]	23.92 (9.14)	19.08 (9.39)	4.85 (3.30 6.39)	<0.001 (0.52)
Hip sagittal range of motion [°]	30.06 (5.72)	27.81 (5.92)	2.25 (0.74 3.76)	0.006 (0.39)
Peak hip extension moment [Nm/kg]	1.52 (0.34)	1.68 (0.36)	-0.15 (-0.26 -0.05)	0.005 (0.44)
Peak hip flexion moment [Nm/kg]	-0.14 (0.19)	-0.10 (0.12)	-0.04 (-0.12 0.03)	0.985 (0.26)
Peak backward trunk lean [°]	0.71 (3.71)	-0.38 (4.44)	1.08 (-0.46 2.63)	0.159 (0.26)
Peak forward trunk lean [°]	5.48 (3.72)	4.12 (4.49)	1.35 (-0.23 2.93)	0.089 (0.32)
Peak pelvis posterior tilt [°]	0.76 (4.50)	-1.29 (4.49)	2.05 (0.99 3.10)	0.001 (0.46)

Table 10.2: Spatiotemporal, hip, pelvis, and trunk parameters during stair descent at baseline and after one week of brace application (braced).

Parameter during stair descent	Baseline	Braced	Mean difference	t-test/WCX
	Mean (SD)	Mean (SD)	(95%CI)	p (d)
Peak pelvis anterior tilt [°]	4.85 (4.48)	3.43 (4.02)	1.43 (0.41 2.44)	0.008 (0.33)
Pelvis sagittal range of motion [°]	4.10 (1.34)	4.72 (1.77)	-0.62 (-1.32 0.08)	0.078 (0.39)
Peak hip adduction [°]	-7.02 (2.68)	-7.35 (2.95)	0.33 (-0.54 1.19)	0.441 (0.11)
Peak hip abduction [°]	4.51 (2.80)	4.20 (3.04)	0.32 (-0.63 1.26)	0.493 (0.11)
Hip frontal range of motion [°]	11.54 (3.85)	11.55 (3.91)	-0.01 (-1.23 1.21)	0.475 (0.00)
Peak hip adduction moment [Nm/kg]	-0.35 (0.23)	-0.51 (0.48)	0.16 (-0.03 0.35)	0.083 (0.39)
Peak hip abduction moment [Nm/kg]	0.56 (0.28)	0.50 (0.34)	0.06 (-0.08 0.20)	0.360 (0.20)
Peak contralateral trunk lean [°]	-1.71 (1.25)	-1.41 (1.49)	-0.30 (-0.94 0.35)	0.349 (0.21)
Peak ipsilateral trunk lean [°]	1.43 (1.48)	1.47 (1.71)	-0.04 (-0.72 0.64)	0.902 (0.03)
Peak ipsilateral pelvis drop [°]	-3.27 (1.83)	-3.86 (1.92)	0.59 (-0.26 1.44)	0.164 (0.31)
Peak ipsilateral pelvis rise [°]	3.79 (1.98)	4.00 (2.49)	-0.21 (-1.07 0.65)	0.617 (0.09)
Pelvis frontal range of motion [°]	7.06 (2.24)	7.86 (3.00)	-0.80 (-2.07 0.47)	0.203 (0.30)
Peak hip int. rotation [°]	-3.22 (9.56)	0.56 (10.14)	-3.79 (-7.63 0.05)	0.053 (0.38)
Peak hip ext. rotation [°]	10.13 (10.51)	11.30 (9.78)	-1.17 (-4.73 2.39)	0.499 (0.11)
Hip transverse range of motion [°]	13.36 (4.60)	10.74 (3.22)	2.62 (0.74 4.49)	0.009 (0.64)
Peak hip internal rotation moment [Nm/kg]	-0.19 (0.12)	-0.17 (0.14)	-0.02 (-0.08 0.04)	0.216 (0.15)
Peak hip external rotation moment [Nm/kg]	0.23 (0.10)	0.29 (0.12)	-0.06 (-0.12 0.00)	0.037 (0.56)
Pelvis transverse range of motion [°]	8.93 (3.07)	6.74 (2.54)	2.19 (1.42 2.96)	<0.001 (0.74)
Stance phase duration [s]	0.70 (0.13)	0.66 (0.10)	0.04 (0.01 0.06)	0.013 (0.29)
Gait cycle duration [s]	1.09 (0.19)	1.03 (0.14)	0.06 (0.01 0.10)	0.020 (0.31)
Mean horizontal CoM velocity [m/s]	0.53 (0.09)	0.56 (0.09)	-0.03 (-0.05 -0.01)	0.016 (0.30)
Mean vertical CoM velocity [m/s]	0.32 (0.06)	0.34 (0.06)	-0.02 (-0.03 0.00)	0.024 (0.29)

Note. External hip moments. P-values as revealed by dependent sample t-tests/Wilcoxon tests (WCX) and effect sizes (|d|); significant results ($p < 0.05$) in bold. 95%CI = 95 % confidence interval.

10.4.1 Effects of hip bracing on pain perception

Participants wore the brace for an average of $10.1 (\pm 3.6)$ hours each day. The average pain score during stair walking decreased from $24.9 (\pm 17.6)$ at baseline to $17.9 (\pm 19.7)$ in the intervention period ($MD = -7.0$; $p = 0.008$). Increases in mean pain scores of approximately 42 % were observed for three participants, while decreases in mean pain scores of approximately 40 % were observed for the remaining 17 participants.

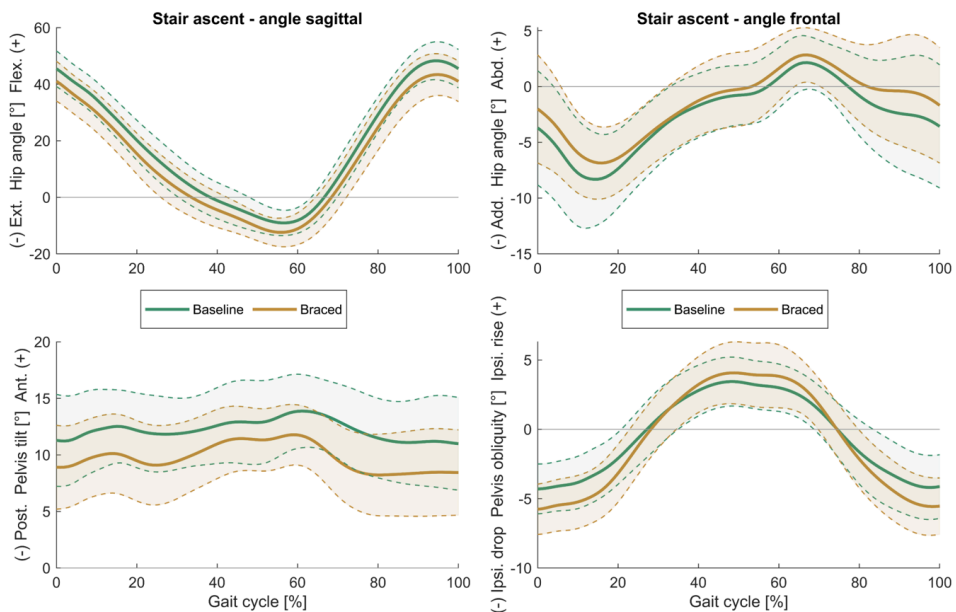


Figure 10.2: Sagittal and frontal plane hip and pelvis angles during stair ascent.

Note. Data presented as mean $\pm$ SD.

10.4.2 Effects of hip bracing on movement velocity

During stair ascent and descent, bracing increased the horizontal (ascent: $MD = -0.02$ m/s, $p = 0.021$; descent: $MD = -0.03$ m/s, $p = 0.016$) and vertical CoM velocity (ascent: $MD = -0.01$ m/s, $p = 0.021$; descent: $MD = -0.02$ m/s, $p = 0.024$) and reduced ST (ascent: $MD = 0.05$ s, $p = 0.005$; descent: $MD = 0.04$ s, $p = 0.013$) and GC duration (ascent: $MD = 0.06$ s, $p = 0.010$; descent: $MD = 0.06$ s, $p = 0.020$).

10.4.3 Effects of hip bracing on hip biomechanics

During stair ascent, bracing increased the peak hip extension angle (MD = 3.27°, $p = 0.003$) and decreased the peak hip flexion angle (MD = 4.82°, $p < 0.001$), peak hip adduction angle (MD = -1.60°, $p = 0.007$) and peak internal rotation angle (MD = -3.67°, $p = 0.039$).

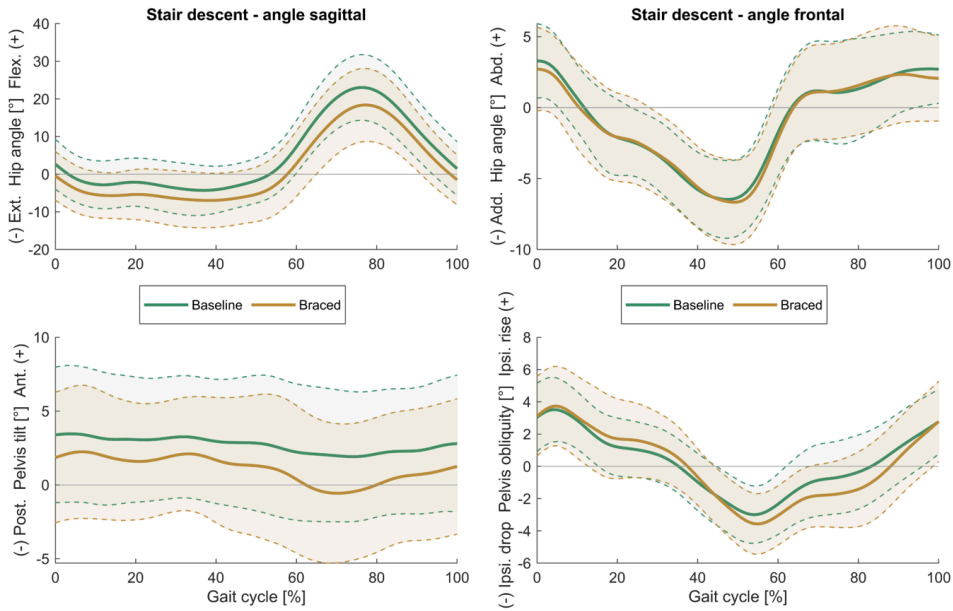


Figure 10.3: Sagittal and frontal plane hip and pelvis angles during stair descent.

Note. Data presented as mean $\pm$ SD.

During stair descent, bracing decreased the peak hip flexion angle (MD = 4.85°, $p < 0.001$) and sagittal and transverse hip RoM (sagittal: MD = 2.25°, $p = 0.006$; transverse: MD = 2.62°, $p = 0.009$), and increased the peak hip extension angle (MD = 2.60°, $p = 0.009$), peak hip extension moment (MD = -0.15 Nm/kg, $p = 0.005$) and the peak external rotation moment (MD = -0.06 Nm/kg, $p = 0.037$).

10.4.4 Effects of hip bracing on pelvis and trunk motion

During stair ascent, bracing increased peak pelvis posterior tilt angle (MD = -2.96° , $p < 0.001$) and decreased peak pelvis anterior tilt angle (MD = 1.95° , $p = 0.007$), resulting in an increased sagittal pelvis RoM (MD = -1.01° , $p = 0.020$). Moreover, bracing increased peak ipsilateral pelvis rise and drop (rise: MD = -0.79° , $p = 0.040$; drop: MD = 1.31° , $p = 0.002$), resulting in a significant increase of frontal pelvis RoM (MD = -2.10° , $p = 0.001$).

During stair descent, bracing increased the peak posterior pelvis tilt (MD = 2.05° , $p = 0.001$) and decreased the peak anterior pelvis tilt (MD = 1.43° , $p = 0.008$) and the transverse pelvis RoM (MD = 2.19° , $p < 0.001$).

Bracing did not influence trunk motion during stair ascent or descent.

10.4.5 Correlation of changes in stair walking biomechanics and pain perception

Table 10.3 and Table 10.4 present the correlations between pain perception and biomechanical parameters for stair ascent and descent, respectively. Scatter plots can be found in the Appendix (Figure A.36, Figure A.37).

None of the biomechanical factors affected by brace application showed a significant correlation with the change in pain perception. However, four parameters during stair ascent (Table 10.3, Figure A.36) and two during stair descent (Table 10.4, Figure A.37) showed moderate correlation coefficients. Decreases in pain score were moderately correlated with decreased pelvis peak ipsilateral rise angle ($r = 0.39$), decreased pelvis frontal ($r = 0.38$) and sagittal RoM ($r = 0.36$), as well as decreased peak hip internal rotation angle ($r = -0.33$) during stair ascent.

Table 10.3: Correlation between change in perceived pain during stair walking and change in biomechanical stair ascent parameters.

Parameter during stair ascent	Pearson's r	95%CI	p
Pelvis peak ipsilateral rise [°]	0.39	-0.06 0.71	0.090
Pelvis frontal range of motion [°]	0.38	-0.08 0.70	0.100
Pelvis sagittal range of motion [°]	0.36	-0.09 0.70	0.114

Table 10.3: Correlation between change in perceived pain during stair walking and change in biomechanical stair ascent parameters.

Parameter during stair ascent	Pearson's r	95%CI	p
Hip peak int. rotation [°]	-0.33	-0.67 0.13	0.155
Hip peak extension angle [°]	0.29	-0.17 0.65	0.211
Hip peak adduction angle [°]	0.22	-0.25 0.60	0.348
Pelvis peak ipsilateral drop [°]	-0.15	-0.56 0.31	0.527
Pelvis peak posterior tilt [°]	-0.12	-0.54 0.34	0.610
Pelvis peak anterior tilt [°]	0.10	-0.36 0.52	0.663
Mean horizontal CoM velocity [m/s]	0.06	-0.39 0.49	0.795
Hip peak flexion angle [°]	-0.03	-0.47 0.42	0.902
Stance phase duration [s]	0.02	-0.43 0.46	0.942
Mean vertical CoM velocity [m/s]	-0.01	-0.45 0.44	0.980
Gait cycle duration [s]	0.00	-0.44 0.44	0.992

Note. 95%CI = 95 % confidence interval; CoM = centre of mass

For stair descent, increased hip peak extension angle ($r = 0.31$) and increased hip transverse RoM ($r = -0.41$) were moderately correlated with decreases in pain score.

Table 10.4: Correlation between change in perceived pain during stair walking and change in biomechanical stair descent parameters.

Parameter during stair descent	Pearson's r	95%CI	p
Hip transverse range of motion [°]	-0.41	-0.72 0.04	0.071
Hip peak extension [°]	0.31	-0.15 0.66	0.177
Hip sagittal range of motion [°]	-0.27	-0.64 0.20	0.249
Pelvis transverse range of motion [°]	-0.25	-0.62 0.22	0.297
Mean horizontal CoM velocity [m/s]	0.24	-0.22 0.62	0.311
Mean vertical CoM velocity [m/s]	0.24	-0.23 0.62	0.299
Pelvis peak posterior tilt [°]	-0.19	-0.59 0.27	0.415
Gait cycle duration [s]	-0.15	-0.56 0.31	0.515
Stance phase duration [s]	-0.14	-0.55 0.33	0.567

Table 10.4: Correlation between change in perceived pain during stair walking and change in biomechanical stair descent parameters.

Parameter during stair descent	Pearson's r	95%CI	p
Hip peak ext. extension moment [Nm/kg]	0.13	-0.33 0.54	0.580
Hip peak flexion [°]	0.12	-0.34 0.53	0.616
Pelvis peak anterior tilt [°]	0.07	-0.38 0.50	0.764
Hip peak ext. external rotation moment [Nm/kg]	0.03	-0.42 0.46	0.910

Note. 95%CI = 95 % confidence interval; CoM = centre of mass

10.5 Discussion

The present study aimed to assess the effects of one week of hip bracing on hip, pelvis, and trunk biomechanics, as well as pain perception, during stair walking in individuals with mild-to-moderate HOA. We found that bracing reduced hip pain during stair walking by approximately 28 %. Biomechanical analysis showed that, in line with our hypothesis, bracing increased movement velocity during both stair ascent and descent, and reduced the peak internal hip rotation angle during stair ascent. In contrast to our hypothesis, limited frontal hip mobility in individuals with HOA was aggravated by brace use, with reductions in peak hip adduction angle during stair ascent. Lastly, bracing did not result in an increase in peak hip internal rotation moment or a decrease in ipsilateral trunk lean, as expected.

10.5.1 Bracing increased movement velocity and hip joint moments during stair descent

Decreased stair walking velocity has been observed in individuals with HOA (Hall et al., 2017), and bracing was able to reverse this effect. During steady-state stair walking, only minor modifications of step length are possible due to the dimensions of the step. Thus, increased velocity was caused by an increase in cadence, as indicated by lower GC durations. Stair descent requires substantial control of the downward movement of the CoM. Increased muscular co-contraction to stiffen the limb in single support has been described as a strategy to reduce vertical CoM velocity and, as a result, reduce the muscular demand for CoM deceleration and the joint contact forces at IC (Buckley et al., 2013). Therefore, increasing the vertical CoM velocity with bracing likely required greater muscle activity to decelerate the CoM and

potentially increased the hip contact force. Thereby, it has been found that the activity of the periarticular musculature is important for shock absorption (Brandt et al., 2008), and greater GMed activation was associated with reduced impact forces during a step-down task in individuals with HOA (Dwyer et al., 2013). Thus, alterations in GMed activity by bracing might mediate between increased contact force due to increased CoM velocity and decreased contact force due to periarticular muscle activity. Additionally, the GMed has been found to contribute to the forward acceleration of the CoM during stair descent (Lin et al., 2015) and thus enhanced GMed activation could also have caused the observed increase in anteroposterior CoM velocity. However, since muscle activity was not captured during this study, further investigation is warranted.

The increased movement velocity likely contributed to the increase in peak hip extension and external rotation moments observed during stair descent, as gait velocity correlates with hip joint kinematics and dynamics (Fukuchi et al., 2019; Lelas et al., 2003). Reduced peak hip extension and external rotation moments have previously been described in individuals with HOA during stair descent (Hall et al., 2017; Meyer et al., 2016); bracing thus counteracted these effects of HOA on stair descent dynamics. However, larger hip extension moments require larger activity of the hip flexors, which are weakened in individuals with mild-to-moderate HOA (Loureiro et al., 2018). Thus, increasing velocity and peak hip extension moments during stair descent might promote muscle fatigue, but simultaneously exercise hip flexor muscles. Future studies should address the long-term effect of hip brace use on the force capacity of the hip muscles.

10.5.2 Bracing reduced hip internal rotation and adduction, but increased pelvis frontal mobility

Although the brace applied in the current study did not aim to alter the femur position mechanically, it reduced peak hip internal rotation during stair ascent by 3.7° and reduced the transverse hip RoM during stair descent by 2.6° . Thereby, reduced hip internal rotation angles during stair ascent were moderately correlated to reduced pain during stair walking. In contrast, a moderate correlation was found between decreased transverse hip RoM during stair descent and increased hip pain. Other braces intending to mechanically externally rotate the hip joint resulted in higher reductions of peak hip internal rotation angles of 6.7° during level walking in individuals with HOA (Nérot & Nicholls, 2017) and of 4.2° and 6.4° during stair ascent and descent, respectively, in individuals with femoroacetabular impingement (Newcomb et al., 2018). However, in those studies, no effect on pain perception was found

(Newcomb et al., 2018), or mixed results were reported (Nérot & Nicholls, 2017). Preventing RoM endpoints from being reached is thought to reduce hip pain (Reiman et al., 2020; St-Pierre et al., 2022). However, internal rotation angles observed during level walking and stair walking likely do not require the entire available RoM, as larger peak angles have been found, e.g. in curved walking (Steingrebe et al., 2024). Therefore, future studies are required to clarify the effect of hip internal rotation on pain in individuals with HOA. As results differ between movement tasks, analyses should focus on the situation in which peak internal rotation occurs, e.g. in combination with hip adduction, or the simultaneous loading condition.

Furthermore, brace application led to a decrease in peak adduction of 1.6° during stair ascent. Similar results have previously been reported, with reductions of 1.9° during level walking (Nérot & Nicholls, 2017) and approximately 3° in stair walking (Newcomb et al., 2018). Limiting hip adduction has been identified as a strategy to increase mediolateral stability and reduce hip contact forces (Wesseling et al., 2015); however, this approach also decreases the demand on the hip abductor muscles, potentially leading to muscle weakness in the long term (Meyer et al., 2018). Despite the reduction in peak adduction, we did not observe a decrease in the external adduction moment, which is closely linked to hip contact forces (Wesseling et al., 2015), either due to the simultaneous increase in movement velocity or a change in frontal pelvis mobility.

Bracing in our study led to an increase in frontal pelvis motion with increased ipsilateral pelvis rise and drop. Although reduced frontal pelvis RoM has been found during level walking in individuals with various degrees of HOA (Bolink et al., 2015; Constantinou et al., 2017; Reininga et al., 2012; Thurston, 1985), analyses of stair walking did not yield any differences from healthy controls (Hall et al., 2017; Meyer et al., 2016). However, in contrast to level walking, stair ascent requires greater concentric effort from the abductor muscles to elevate the pelvis on the contralateral side, facilitating step clearance of the swinging leg (Nadeau et al., 2003). Increased oblique pelvis motion observed with bracing may result from increased SIJ mobility, facilitating step clearance during stair ascent, which could ultimately lead to a lower risk of falling (Templer, 1995). Furthermore, reducing the contralateral pelvis drop during level walking can increase hip joint contact forces by 11.9 % (Diamond et al., 2022), resulting in decreased hip pain. Peak ipsilateral pelvis rise angle and frontal pelvis RoM showed the strongest correlation with change in pain score in our study (Table A.8; Figure A.36), and larger increases in pelvis rise and frontal RoM led to larger increases in pain. Thus, the increased frontal pelvis mobility, especially the increased contralateral pelvis drop observed with bracing, seems to be detrimental to the improvements in hip pain.

Lastly, it remains unclear whether the reductions in peak hip internal rotation and adduction angles are due to the passive resistance of the hip brace or result from an altered movement strategy and muscle activation pattern. As the brace did not affect peak hip adduction angle during stair descent and level walking (Steingrebe et al., 2022), passive resistance seems less likely. However, the observed peak hip adduction angles were greatest during stair ascent. Since the brace applies a friction massage to the GMed above the greater trochanter, altered activity of the GMed could have caused the observed changes in hip adduction and pelvis obliquity. The GMed not only counteracts the hip adduction moment but also contributes to external rotation of the hip joint in phases of low hip flexion and to internal rotation of the hip joint in phases of high hip flexion (Delp et al., 1999). Therefore, altered activity of the GMed could also have impacted peak hip rotation.

10.5.3 Bracing altered pelvis tilt and shifted the hip motion towards increased hip extension

A typical posture in individuals with HOA is characterised by lumbar lordosis and increased anterior pelvis tilt (Tateuchi, 2019), which is a compensatory mechanism for reduced hip extension ability (Thurston, 1985). Hip bracing in our study led to a reduction in anterior pelvis tilt, thereby counteracting the postural adaptation described for individuals with unilateral (Thurston, 1985) and bilateral moderate-to-severe HOA (Kubota et al., 2007; van Dronkelen, Braun, et al., 2021). Furthermore, brace application shifted the sagittal peak hip angles towards decreased flexion and increased extension. As the hip joint angle is calculated by the motion of the femur relative to the pelvis, this shift in hip angles is likely caused by the more upright pelvis position, which results in a more extended position of the hip joint without altering the femur's position in global space. Nevertheless, decreased peak hip extension is a prominent feature of HOA gait, becoming more distinct with increasing severity of HOA (Steingrebe et al., 2023). However, bracing did not increase peak hip extension during level walking (Nérot & Nicholls, 2017; Steingrebe et al., 2022), where higher hip extension is required due to the longer step length.

Additionally, bracing reduced peak flexion by 4.8°, which is comparable to the 5.3° previously reported in individuals with femoroacetabular impingement during stair walking (Newcomb et al., 2018). Limiting peak hip flexion during stair walking may be beneficial, as greater hip flexion angles during level walking (Kumar et al., 2018) and stair walking (L. Chen et al., 2024) have been associated with faster disease progression.

Decreased hip extension at terminal stance is linked to a reduced hip contact force and a shift in force direction (Wesseling et al., 2015). Consequently, increases in peak hip extension likely caused an increase in hip contact force. For KOA, it has been shown that excessive knee loading drives KOA progression (Saxby & Lloyd, 2017); however, results for HOA are equivocal. Modifications in gait patterns of individuals with HOA have often been linked to reduced joint loading and pain avoidance (Meyer et al., 2018), and high peak contact hip stress has been found to increase the risk of hip replacement surgery (Recnik et al., 2009). However, gait interventions that increase hip contact forces have been associated with reduced hip pain (Diamond et al., 2022). Likewise, in our study, the observed increases in hip moments and hip peak extension might have led to increased hip contact forces, but were accompanied by a reduction in hip pain. Thereby, increased peak hip extension was correlated to decreased pain perception during stair descent (Table A.9; Figure A.37). Therefore, future studies should clarify the impact of hip bracing on hip contact forces, as an optimal amount of hip loading may be beneficial in reducing hip pain.

10.5.4 Strengths and limitations

Our study is the first to assess the effects of one week of hip bracing on pain perception and biomechanics during stair walking in individuals with HOA. Using a combination of radiographic and functional HOA assessments allowed us to ensure that participants are representative of the target group for conservative treatment methods, as opposed to other studies that included participants with end-stage HOA (Nérot & Nicholls, 2017). Furthermore, the higher wear comfort of elastic hip braces, as applied in the current study, likely increases therapy adherence compared to braces constructed to unload the hip joint, which are often reported to be uncomfortable (Eyles et al., 2022; Kemker et al., 2021; Newcomb et al., 2018).

Furthermore, stair walking biomechanics were assessed at the midpoint of a six-step instrumented staircase and are therefore representative of a steady-state stair walking motion, as opposed to transition steps between level and stair walking, often reported in other studies (Hall et al., 2017; Popovic et al., 2021).

Besides these strengths of our study, some limitations should also be considered. The evaluation of hip and pelvis biomechanics using optical motion capture necessitates the placement of markers on anatomical landmarks. In the braced condition, it was not feasible to attach markers directly to the skin above the anterior and posterior superior iliac spines. Consequently, markers for the anterior spine were placed on the hip belt, while markers for the

posterior spine were affixed through the mesh fabric of the pelvis belt. Applying markers to clothing rather than to the skin increases the risk of relative motion between the anatomical landmark and the marker (Milner, 2008). Alternative methods, such as clusters, are often reported in knee brace analyses (Focke et al., 2020; Turner et al., 2021) but are not applicable for the hip joint as the brace entirely covers the pelvis, and the adjoining segments (thighs and torso) exhibit substantial portions of wobbling mass. However, as the pelvis belt of the brace was fitted very tightly and stair walking movements are relatively slow in comparison to running or jumping, relative movement between the pelvis and the brace is likely to be minimal, although it cannot be entirely ruled out.

The observed changes in hip kinematics induced by bracing were 2.6° to 4.8° in the sagittal plane, 1.6° in the frontal plane, and 3.7° in the transverse plane. Small changes always raise questions about their clinical relevance. However, bracing also induced reductions in pain perception, and some parameters were moderately correlated with this change. Thus, while the absolute changes were small, they might still have affected pain perception. Overall, interpreting transverse plane biomechanics requires caution, as marker-based motion capturing is susceptible to errors in this plane (Leardini et al., 2005). For hip biomechanics, accurately estimating the HJC is crucial (Stagni et al., 2000). Therefore, the equations of Harrington et al. (Harrington et al., 2007) were used for HJC estimation in our study, as they are the most accurate regression method (Kainz et al., 2015).

Furthermore, it was not possible to blind participants to the bracing condition, which raised the risk of placebo effects, especially in subjective parameters such as pain perception.

Lastly, the effect of bracing was evaluated after only one week of wearing the brace. As bracing is likely to be used over a more extended period in the conservative treatment of HOA, the long-term effects must be evaluated before recommendations for or against its use can be made.

10.6 Conclusion

We examined the effect of hip bracing on hip pain and biomechanics of the hip, pelvis, and trunk during stair walking in individuals with mild-to-moderate HOA. Bracing reduced hip pain and reversed some of the effects of HOA on biomechanics by increasing movement velocity, peak hip extension angle, and peak hip extension and external rotation moments, while decreasing anterior pelvis tilt. Thereby, reductions in hip internal rotation during stair

ascent and increases in hip extension during stair descent were moderately correlated with reductions in pain score. In contrast, hip bracing exacerbated the reduction in hip adduction observed in individuals with HOA and induced additional changes, such as increased oblique pelvis motion. As the brace used in this study primarily comprised elastic components and targets soft tissue around the pelvis, hip and ISJ, future studies should further evaluate the underlying adaptation in muscle activation that led to the observed changes in stair walking kinematics and dynamics. Furthermore, these modifications likely increased hip contact forces, which should be evaluated in the future, especially with regard to the long-term effects of hip bracing.

All in all, our results indicate that hip bracing is effective in reducing hip pain, enhancing posture and improving hip function during the demanding task of stair walking. Therefore, bracing should be considered in the conservative treatment of mild-to-moderate HOA as a low-cost option that allows patients to self-manage their therapy.

11 Discussion

11.1 Aim and Main Findings

The aim of this thesis was to investigate the effects of mild-to-moderate HOA on locomotion and, subsequently, to evaluate the effectiveness of hip bracing for treatment. This was achieved by combining biomechanical motion analysis of various locomotion tasks with assessments of pain, hip proprioception, and functional capacity to gain a comprehensive understanding of disease features and treatment effects.

In summary, this thesis yielded the following findings:

1. Individuals with mild-to-moderate HOA have lower functional capacity in the 6MWT.
2. Individuals with mild-to-moderate HOA have altered locomotion biomechanics, namely lower hip RoM in all movement planes caused by limited hip extension, hip adduction and hip internal rotation, lower hip moments and altered sagittal knee and pelvis motion.
3. During stair walking, individuals with mild-to-moderate HOA adopt a movement strategy that reduces the risk of falling, albeit at the cost of reduced gait efficiency.
4. Movement tasks have to be more demanding to detect mobility limitations in individuals with mild-to-moderate HOA than in individuals with more severe HOA. Thereby, limitations in sagittal hip mobility are most noticeable during level walking, in frontal hip mobility during spin turning and stair ascent and in transverse hip mobility during the step turn.
5. Mild-to-moderate HOA or hip bracing do not impact hip joint position sense. More sensitive measurement methods are needed to assess subtle changes in hip joint position sense.
6. Hip bracing for one week reduced hip pain during walking, stair walking and at rest (night pain), and increased locomotion velocity, consequently improving functional capacity as measured using the 6MWT.

7. Bracing alters sagittal hip and pelvis motion during level walking, indicating improved pelvis mobility, likely responsible for the observed increase in locomotion velocity.
8. Bracing with an elastic hip brace reduces peak hip adduction and internal rotation angles during stair ascent and step turning, potentially reducing hip pain, joint loading and muscle activity.
9. Bracing increases hip extension and external rotation moments during level walking and stair descent, as well as knee abduction and external rotation moments during the spin turn.

The effects of mild-to moderate HOA and bracing on hip joint kinematics and dynamics during the various locomotion tasks are summarised in Figure 11.1 and Figure 11.2.

11.2 Effects of hip osteoarthritis on proprioception, postural control, functional capacity and locomotion biomechanics

Understanding the various implications of mild-to-moderate HOA on locomotion is essential for assessing the effectiveness of treatments, such as bracing (Broström et al., 2012). To gain a comprehensive understanding of these implications, alterations in locomotion biomechanics and resultant limitations in functional capacity were studied. Furthermore, as one mechanism proposed to explain the beneficial effects of elastic braces is the improvement of proprioception (Cudejko et al., 2018), a JPR test was included in our study to measure hip flexion JPS. We found no difference in the absolute error during the JPR test between individuals with and without HOA. This contradicts findings from individuals with KOA for whom a decreased JPS has been frequently shown (Knoop et al., 2011), as well as the findings of one study including individuals with HOA (Alkhamis et al., 2024). However, individuals in the latter study showed considerably larger errors during the reproduction of hip flexion and abduction angles than those found in the present study. While a comparable HOOS of the participants in the present study and the study of Alkhamis et al. (2024) suggests a similar level of HOA severity, the participants in the present study reported less pain in the hip joint (25.7 mm vs. 54.76 mm on a 100 mm VAS), and pain has been shown to impact proprioception (Efsthathiou et al., 2022).

Additionally, a decline in hip rotation JPS with age has been found in one study (Wingert et al., 2014); however, in another study by Pickard et al. (2003) no age effect on frontal hip JPS was found. The authors suggested that the high level of physical activity in their cohort of older adults could explain the lack of difference, as higher repositioning errors have been found in sedentary older adults compared to active older adults (Petrella et al., 1997). Individuals with HOA in the present study were still physically active, as indicated by a Tegner Activity Score of 4.7 ± 0.8 (Tegner & Lysholm, 1985), which is consistent with reference data from this age group (Briggs et al., 2009). This suggests that mild-to-moderate HOA, associated with low levels of pain combined with sufficient physical activity, does not or only minimally impair proprioceptive perception.

Another explanation for the observed results might be that the measurement technique used in this study was unable to detect any such differences, as we found a minimal detectable change of about 2° in the applied JPR test. According to Hillier et al. (2015), active JPR tests, as applied in the current study, reflect the functional use of proprioception but require sufficient kinesthetic memory of the pre-established position and motor control to readopt the joint position. As hip flexion movement in our study was performed while standing on one leg, maintaining balance and simultaneously flexing the hip joint against gravity likely required substantial muscle strength and motor control. Therefore, limitations in strength and motor control observed in individuals with HOA (M. M. Arokoski et al., 2002; Higgs et al., 2023; Judd et al., 2014; Loureiro et al., 2013) might have masked underlying differences in the sensory quality of hip proprioception. While performing the movement in a standing position reflects the position of the joint during locomotion tasks, conducting the JPR test while supine, as done by Alkhamis et al. (2024), might reduce this interference.

Lastly, as discussed initially (chapter 2.1), it is necessary to distinguish between illness and disease when defining HOA (Kraus et al., 2015). While we used a combination of radiography, pain, and functional characteristics to define the inclusion criteria of the intervention group, we could, in the majority of cases, only rely on functional parameters, and thus the absence of illness, for the inclusion of control participants. Thus, it cannot be ruled out that individuals in the control group had signs of HOA disease at the molecular or anatomic level affecting hip proprioception.

Several studies have shown a relationship between proprioception and postural control (Alkhamis et al., 2024; Wang & Fu, 2022; Wingert et al., 2014). We studied the effect of HOA on postural control during stair walking, as maintaining dynamic balance is highly demanding during this task due to the restriction of the base of support by the dimensions of the stair tread (Novak et al., 2016). Thereby, dynamic balance was assessed using the inverted

pendulum model with an extrapolated CoM and quantified by the MoS (Hof, 2008; Hof et al., 2005). Although we did not find an effect of HOA on hip proprioception, individuals with HOA ascended stairs with a larger anterior MoS than individuals without HOA. While not significant, the same tendency was seen during stair descent. Thus, individuals with HOA use a less unstable movement strategy when walking stairs. Similar results have been reported for individuals with KOA, where higher performance in the TUG was correlated with a lower anterior MoS during a step-down movement (Koyama et al., 2015). However, a negative anterior MoS is necessary for an energetically efficient transition to the next step (Kuo et al., 2005). Therefore, individuals with HOA demonstrate a movement strategy that reduces the risk of falling, albeit at the cost of reduced gait efficiency (Bosse et al., 2012).

Besides hip proprioception and dynamic balance, functional capacity was also assessed in our study using the 6MWT. Individuals with HOA had a decreased performance by 11.4 %. This result aligns with another study involving participants with mild-to-moderate HOA, reporting decreased performance of 10.6 % (Rydevik et al., 2010), and is lower than the reductions of 28 % reported for individuals with end-stage HOA (Judd et al., 2014). However, the absolute distances covered during the 6MWT by both groups of 560 and 632 m in this study were lower than the previously described 628-673 m for individuals with mild-to-moderate HOA and 719 m for healthy controls (Eitzen et al., 2015; Kumar et al., 2015; Rydevik et al., 2010). These differences could either stem from different test settings with a circuit of 54 m length including four 90° turns in our study and a corridor of 20 m length including two 180° turns in the studies of Eitzen et al. (2015) and Rydevik et al. (2010), or they could stem from a diverging definition of HOA based on radiographic imaging but not symptoms (Kumar et al., 2015). Nevertheless, our study confirmed that functional capacity is already decreased in individuals with mild-to-moderate symptomatic HOA.

The spatiotemporal gait characteristics of individuals with HOA can explain the observed decrease in distance covered during the 6MWT. In a meta-analysis, lower gait speed, caused by a decrease in step length, and increased limb asymmetries have been described (Constantinou et al., 2014). In line with these results, individuals with HOA in this study showed a decrease in step length of 7.5 % and a lower movement velocity of 6.8 %, 6.0-8.6 % and 3.1-4.2 % during level walking, stair walking and turning, respectively. These differences are smaller than the 26 % reported in the meta-analysis (Constantinou et al., 2014). However, data from individuals with end-stage HOA scheduled to undergo THR were included in their analysis.

While these spatiotemporal gait characteristics are easy to obtain (R. Davis, 1997; Kirtley, 2006), they contain only limited information on the underlying causes. By analysing joint kinematics and movement patterns, the underlying causes for changes in functional capacity and spatiotemporal gait characteristics can be explored. To summarise the current knowledge on the impacts of HOA on locomotion biomechanics, we conducted a systematic review and meta-analysis of existing literature on kinematic changes in locomotion tasks observed in individuals with HOA.

The review showed that most research focused on straight level walking, a few studies analysed stair walking, and one study analysed curved walking movements. Results from our data showed that different movements require different hip mobility, evoke different levels of hip moments (Figure 11.1), and that differences in the movement patterns between individuals with and without HOA depend on the evaluated movement task. During level walking, the meta-analysis yielded kinematic differences in the ipsi- and contralateral hip and knee joints of individuals with HOA. While changes at the hip occurred in all motion planes, changes in knee kinematics occurred mainly in the sagittal plane. Pooling of data across studies was challenging due to the large heterogeneity in the evaluated parameters. However, the most prominent kinematic feature across multiple studies was a lower sagittal hip RoM caused by lower hip extension angles. Most previous studies assessed kinematic changes induced by HOA in populations with severe HOA shortly before surgery. Limited availability of data from individuals with less severe HOA impedes the analysis of severity effects. Nevertheless, the meta-analysis showed that sagittal hip RoM and peak hip extension are more restricted in individuals with severe than in those with mild-to-moderate HOA. In our cohort of individuals with mild-to-moderate symptomatic unilateral HOA, we also found lower sagittal hip RoM caused by lower peak hip extension compared to the controls during straight walking and turning, but not during stair walking. This coincides with the peak hip extension angles being larger in level walking and turning and is in line with results from other studies, including individuals with mild-to-moderate HOA on level walking (Constantinou et al., 2017; Leigh et al., 2016). Moreover, in line with the results from the meta-analysis on gait kinematics, lower peak knee flexion was observed during the step turn.

Although the meta-analysis did not find significant differences in ankle or pelvis kinematics, results from individual studies still suggest the presence of altered kinematics, e.g. increased anterior pelvis tilt (Kubota et al., 2007), in certain subject groups and settings. Individuals from our study also exhibited increased pelvic tilt RoM during level walking compared to the control group. Similar results were found by Watelain et al. (2001) in a group of people with

mild HOA. Thereby, anterior pelvic tilting has been proposed as a compensatory mechanism for reduced hip extension mobility (Lee et al., 1997).

Furthermore, the meta-analysis revealed reduced frontal and transverse hip RoM during walking. Although the results are not as uniform as those for the sagittal plane, individual study results indicate that primarily lower peak hip abduction (Kubota et al., 2007; Rutherford et al., 2015a) and adduction (Kataoka et al., 2021; Tateuchi et al., 2014) contribute to decreased frontal hip RoM. In the transverse plane, especially limited hip internal rotation significantly impacts overall mobility (Rutherford et al., 2015a). In our cohort of individuals with mild-to-moderate HOA, we also found lower frontal and transverse hip RoM but only in movement tasks that require substantial hip adduction, such as stair ascent and spin turning, or internal rotation, such as straight walking and turning.

In summary, alterations in spatiotemporal, 3D hip joint and sagittal knee and pelvis kinematics were found in individuals with unilateral mild-to-moderate HOA; however, as the deviations from healthy controls are smaller than in individuals with more severe HOA, movement tasks must be demanding enough to reveal limitations in hip mobility.

Besides the differences observed in joint kinematics, lower hip moments were found in individuals with HOA compared to controls. While lower hip internal rotation moments were found during stair ascent, lower hip flexion, extension, adduction and external rotation moments were found during level walking. Previous studies, including participants with mild-to-moderate HOA, also described lower hip extension moments during level walking (Constantinou et al., 2017), and reduced hip moments have been found, especially with reduced step length (Tateuchi et al., 2021). The reduction in peak adduction moment has also been reported in a meta-analysis of HOA walking dynamics (Diamond et al., 2018). However, the authors suggested that the alterations mainly occur in individuals with end-stage HOA. One reason for this might be that the studies included in the meta-analysis used matched walking speeds or statistical methods to adjust for differences in walking speed between the healthy and the HOA groups. As the hip adduction moment correlates with walking speed (Rutherford & Hubley-Kozey, 2009), this might artificially lower the hip adduction moment observed in healthy controls. However, our results show that joint moments are lower, even in individuals with mild-to-moderate HOA, under natural walking conditions. Thereby, the peak adduction moment is considered a crucial factor for hip joint loading (Wesseling et al., 2015) since it must be resisted by the hip abductor muscles. Tateuchi et al. (2017) found that the cumulative hip moment in the frontal plane, in particular, and possibly the cumulative hip moment in the sagittal plane, can predict the radiographic progression of HOA. Therefore, reduced hip

joint moments likely stem from pain-avoidant gait strategies employed by the affected persons and diminished muscle function (Meyer et al., 2018; Ornetti et al., 2011; Watelain et al., 2001).

Overall, the largest hip joint moments were found during turning, likely caused by high mediolateral GRF (Glaister et al., 2007). However, no group differences in joint moments were found during turning. Thereby, both groups demonstrated large interindividual (Figure 11.1) variability in hip joint moments during these movements, which indicates the existence of individual movement strategies for completion of this movement task. Identification of diverging whole-body movement strategies during turning and their associated hip joint loading could, therefore, serve as a foundation for subsequent gait retraining interventions to modify hip joint loading in individuals with HOA.

In summary, this thesis emphasises the existence of locomotion impairments in terms of altered hip, pelvis and knee biomechanics and lower functional capacity at an early disease stage. This underlines the need for effective conservative treatment options to improve HOA treatment in this early phase.

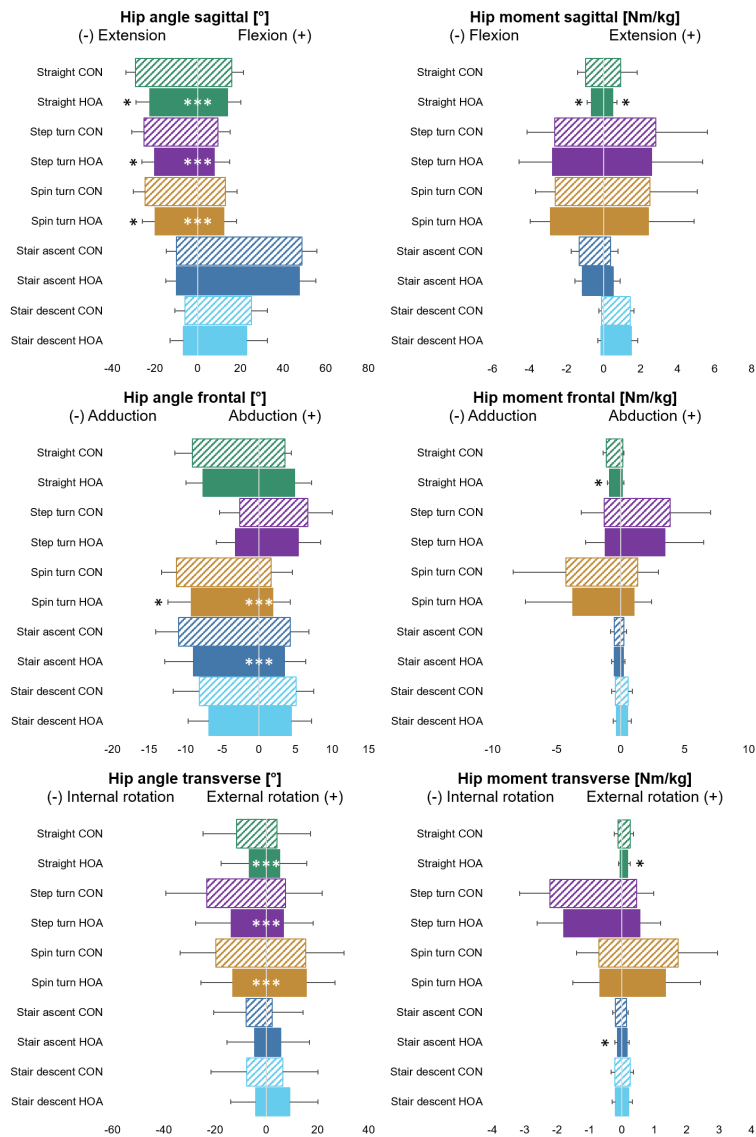


Figure 11.1: Overview of hip osteoarthritis (HOA) effects on hip joint angles and moments.

Note. Data presented as mean $\pm$ SD. CON = control group. * = significant ($p < 0.05$) difference between HOA and CON in peak angles/moments; *** = significant ($p < 0.05$) difference between HOA and CON in range of motion.

11.3 Effect of hip bracing on pain, functional capacity and hip proprioception

This thesis evaluated the effectiveness of an elastic hip brace for the treatment of mild-to-moderate HOA.

The purpose of the applied elastic hip brace was to stabilise the pelvis girdle and SIJ, alleviate pain through trigger point massage of the SIJ and GMed regions, and restore functional hip mobility through continuous friction of GMed. This approach contradicts previous bracing concepts applied in individuals with HOA, which aimed to mechanically reduce or alter loading of the hip joint (Nérot & Nicholls, 2017; T. Sato et al., 2008).

The most prominent and disabling symptom of HOA is joint pain, which is typically the primary reason for consulting a physician (Juhakoski et al., 2008). This thesis demonstrated that one week of brace usage reduces hip pain during walking, stair walking and at rest (night pain). The observed absolute changes in mean pain score were rather small, with 7.3 mm for walking, 7.0 mm for stair walking and 3.1 mm for night pain. These absolute changes represent relative changes of 28.4 % for level walking, 28.2 % for stair walking and 18.2 % for night pain. In individuals with HOA, a MCID of 15.3 mm or 32 % has been proposed (Tubach et al., 2005), and thus, the clinical importance of our finding is uncertain. However, anchor-based methods for detecting MCIDs yield highly variable results, and it has been noted that the MCID is highly subjective, as every individual has to decide which improvement is important enough to undergo treatment (Dekker, 2019). Participants in the present study wore the brace on average 10.1 ± 3.5 hours per day, which is more than twice as long as prescribed (four hours per day), indicating that participants perceived the brace as beneficial and were willing to comply with treatment.

Lastly, it must be considered that the overall treatment effect is likely a combination of the actual treatment effect and additional effects, such as the placebo effect or the regression to the mean effect (Englund, 2023). However, blinding of participants to the treatment condition can not be established for bracing interventions, and thus, the inclusion of a placebo group for estimating the placebo effect size was not feasible. Nevertheless, the described effects of bracing were found after a rather brief intervention period of one week. Other studies have shown a decrease in pain during the initial three months of brace use (T. Sato et al., 2008). Extending the duration of brace use could therefore enlarge the improvement in pain observed after one week of brace wear.

In line with previous findings on brace-induced improvements in the TUG test (E. E. Sato et al., 2012), functional capacity, as measured using the 6MWT, increased on average by 5 % with bracing in our study. Higher physical function in the 6MWT is correlated with higher perceived self-efficacy, which, in turn, impacts health-related quality of life (Sundén et al., 2013). The increase in functional capacity was observed after medium-term but not short-term bracing, indicating that the beneficial effects of bracing on functional capacity evolve with prolonged brace use. Regarding pain, larger benefits were found after long-term brace use in the study of E. Sato et al. (2012). Therefore, extended use of the brace may enhance the beneficial effects on functional capacity seen after using medium-term bracing.

Our study found no impact of bracing on hip proprioception. While this result contradicts research on knee bandages (Baltaci et al., 2011; Beynnon et al., 1999, 2002; Bodendorfer et al., 2019; Selfe et al., 2008, 2011), it aligns with the absence of differences in joint position sense between individuals with and without HOA in our study. Nevertheless, positive effects of knee bracing have also been found in healthy young individuals who are not expected to experience proprioceptive loss (Baltaci et al., 2011). Therefore, it remains unclear whether hip bracing can influence hip joint proprioception. As discussed previously (chapter 11.2), more sensitive testing methods may be needed to detect subtle changes in hip joint position sense induced by HOA or bracing.

11.4 Effect of hip bracing on locomotion biomechanics

Across all analysed movements, bracing led to an increase in movement velocity. However, the mechanism by which the increase in velocity was obtained likely differed between movement tasks. During level walking, step length increased with a simultaneous increase in pelvis tilt and rotation, which may allow individuals with HOA to increase step length despite limitations in hip extension and internal rotation mobility (Leigh et al., 2016; Liang et al., 2014). During stair walking, step length cannot be modified due to the tread dimensions; however, lower ST duration is indicative of increased cadence. Thereby, in contrast to level walking, decreased pelvis rotation was observed during stair descent, while increased pelvis tilting and obliquity motion were found during stair ascent.

Overall, bracing tended to decrease the peak hip flexion angle and increase the hip extension angle (Figure 11.2). This shift in sagittal peak joint angles is likely caused by a less anteriorly tilted position of the pelvis. Increased anterior pelvis tilt has been described as a typical posture in individuals with HOA (Tateuchi, 2019). Thus, bracing might be able to counteract this HOA-induced alteration in pelvis motion. However, significant differences in sagittal peak hip

angles between bracing conditions were only observed during stair walking, which requires greater hip flexion but lower hip extension compared to level walking or turning. Thus, erection of the pelvis by bracing might only be possible during movements that do not require the entire available hip extension range of motion.

As the pelvis is closely connected to the lumbar spine (Ike et al., 2018; Thurston A J & Harris, 1983; Whittle & Levine, 1999), the observed alterations in pelvis motion are likely to affect lumbar spine mobility (Hurwitz et al., 1997; Watelain et al., 2001). Individuals with HOA frequently report lower back pain (Thurston, 1985). While chronic back pain is associated with increased pelvic horizontal rotation (Huang et al., 2011), acute lower back pain is associated with lower pelvic horizontal rotation, potentially to reduce forces transmitted across the lumbar spine (N. Taylor et al., 2004), but increased pelvic and lumbar spine motion is not correlated with back pain in patients with HOA (Thurston, 1985). Thus, the interplay between pelvis and lumbar spine motion and pain remains a topic of controversy. The applied hip brace exerts pressure around the pelvis girdle and onto several trigger points at the SIJ. Future research needs to clarify whether brace-induced increases in pelvic motion are enabled by a stabilisation of the pelvis-spine complex or pain reduction, and whether increased pelvic motion causes greater stress on the lumbar spine, potentially triggering lower back pain with long-term brace use.

Apart from the modification of sagittal hip and pelvis kinematics, changes in peak hip adduction were observed during stair ascent, during which higher peak adduction was observed compared to level walking. Limiting hip adduction has been identified as a strategy to increase mediolateral stability and reduce hip contact forces (Wesseling et al., 2015). However, reducing hip adduction also decreases the demand on the hip abductor muscles, potentially leading to muscle weakness in the long term (Meyer et al., 2018). Hip abductor muscles have been found to be weakened in individuals with HOA (M. M. Arokoski et al., 2002; Burgess et al., 2022; Judd et al., 2014; Loureiro et al., 2013; Marshall et al., 2016). Therefore, long-term evaluation of brace-induced effects on muscle forces, especially of the hip abductor muscles, is warranted.

Furthermore, by limiting peak hip internal rotation, lower transverse hip RoM was observed with bracing during turning and stair descent. Similar peak hip internal rotation angles were found during the spin and step turn in individuals with HOA, while in those without HOA, larger internal rotation angles were observed during the step turn than the spin turn. Thus, both turning conditions may require the entire available transverse hip RoM of the HOA participants. Thus, by decreasing peak internal rotation, bracing may prevent RoM endpoints

from being reached, which is likely associated with hip pain (Reiman et al., 2020; St-Pierre et al., 2022). This aligns with our finding of a correlation between reduced hip internal rotation angles during stair ascent and reduced pain during stair walking. Moreover, pain with internal hip rotation is a key feature of clinical diagnosis of HOA (Altman et al., 1991). The brace applied in the study of Nérot & Nicholls (2017) also aimed to reduce hip internal rotation; however, in contrast to the elastic brace in this study, the brace was designed to mechanically induce abduction and outward rotation of the femur. Despite the lack of mechanically induced external rotation, the elastic hip brace was also capable of restricting excessive hip internal rotation, resulting in beneficial effects on hip pain.

Lastly, it remains unclear whether the brace-induced alterations in joint kinematics stem from mechanical restrictions of the brace or neuromuscular adaptations. Especially limitations in frontal and transverse plane hip kinematics are likely to stem from mechanical restriction of the brace to these movements, as the brace joint only contains one degree of freedom in the sagittal plane.

The inclusion of a retention analysis, in the form of a motion analysis conducted after the bracing interventions but without the application of the brace during measurement, may allow for the analysis of the underlying causes of the brace effects and also the retention of bracing effects over time.

Hip joint moments remained largely unaltered by bracing. Nevertheless, increased hip extension moments were observed during straight-level walking and stair descent (Figure 11.2). Increases in peak hip moments have been linked to increases in gait velocity, especially when achieved by increasing step length (Tateuchi et al., 2021), as found in our study during level walking. During level walking, the peak of the extension moment is reached during terminal stance. A longer step length caused by a more forward-tilted pelvis may increase the distance between the foot (and hence the centre of pressure) and the CoM (Leigh et al., 2016). Thus, the GRF lever arm might be increased, resulting in larger extension moments. However, despite an increase in gait velocity and step length, none of the other hip joint moments increased, and step length cannot be altered during stair descent. In a longitudinal study investigating the effects of a medical foot device in a population of people with HOA over one year, an increase in hip extension moment was accompanied by a decrease in hip pain (Solomonow-Avnon et al., 2017). This finding aligns with results of Hurwitz et al. (1997), who reported a negative correlation between hip pain and peak hip extension moment. Thus, the pain relief caused by brace application might have enabled the HOA patients to increase the hip extension moment. However, larger hip extension moments require greater muscle activity of the hip flexors, which have been shown to be weakened in individuals with mild-

to-moderate HOA (Loureiro et al., 2018). Thus, increasing velocity and peak hip extension moments might promote muscle fatigue but simultaneously exercise weakened hip flexor muscles. Thus, again, future studies should address the long-term effects of hip brace use on the force capacity of the hip muscles.

Apart from a reduction in peak knee flexion during the step turn, knee kinematics and dynamics were not impacted by HOA in our study. However, bracing induced several changes in knee biomechanics, such as reduced peak knee flexion during stair walking or increased frontal and transverse knee moments during stair descent and spin turning. While the increase in the knee external rotation moment during stair descent diminished after medium-term brace use, the changes observed during spin turning remained with medium-term bracing. As joint moments and movement velocity are closely linked (Bejek et al., 2006; Fukuchi et al., 2019; Lelas et al., 2003; Zeni & Higginson, 2009), the observed increases might stem from the observed increase in movement velocity. However, only a few of the analysed peak moments increased with bracing despite an overall increase in movement velocity.

HOA has been linked to an increase in knee moments and knee contact forces, consequently raising the risk of developing KOA (Shakoor et al., 2003; Stief et al., 2018). Therefore, future studies need to further investigate the impact of hip bracing on the knee joint, especially with regard to long-term bracing effects, to clarify potentially detrimental effects of increased knee loading with bracing.

Overall, it has to be considered that the insights from the biomechanical motion analysis conducted in this study are limited to joint angles and joint moments. While these parameters allow a precise description of the observed motion and the resultant external forces acting at the joint, the role of muscular activity in the generation of the observed joint kinematics, as well as its role in joint loading, remains unexplored. However, joint moments have been shown to correlate with joint loading at the hip (Wesseling et al., 2015) and knee joint (Holder et al., 2020; Manal et al., 2015) and are widely used as surrogate measures. Nevertheless, future studies should also analyse muscle activity and apply more complex musculoskeletal models to investigate the effects of bracing on muscle activation and joint loading.

In this study, the effects of bracing on locomotion biomechanics were assessed not only immediately after brace application but also after one week of brace use. Overall, the effects of bracing on locomotion biomechanics were similar for short and medium-term bracing (Figure 11.2). Interestingly, the brace-induced reductions in peak hip flexion and adduction appear to be largest immediately after brace application, whereas the reduction in hip internal rotation increases with brace wear time, indicating different underlying mechanisms.

Studies analysing the effects of knee braces also showed immediate bracing effects on knee kinematics and dynamics (Schween et al., 2015), and brace effects on knee loading did not change over time (Peebles et al., 2019). In contrast to the immediate effects of bracing on locomotion biomechanics, the beneficial effect on functional capacity evolved only after medium-term bracing. Nérot & Nicholls (2017) reported an immediate reduction in hip pain with bracing in nine out of 14 participants, while T. Sato et al. (2008) showed increases in pain reduction and functional capacity with prolonged brace use; however, in the course of three months rather than one week as in the current study.

In summary, brace effects such as stabilisation, change in skin temperature or compression (Cudejko et al., 2018) immediately influence locomotion biomechanics, while functional capacity increases with medium-term brace use. Analyses of both the immediate and long-term effects of bracing on pain are needed to understand the chronology and interplay of brace-induced changes in joint biomechanics, pain and functional capacity.

While bracing was able to “normalise” locomotion in some aspects, such as the increase in locomotion velocity, the erection of the pelvis and simultaneous increase in hip extension during stair walking, or the increase in hip joint moments, other HOA-induced locomotion features, such as reduced hip adduction or internal rotation, remained unaltered or were even engraved by bracing. This finding aligns with results from patients treated with THR. Beaulieu et al. (2010) showed that alterations in gait biomechanics, such as reduced peak flexion angle, peak extension angle and sagittal plane RoM, remained even ten months after surgery. Similar results were found by Foucher et al. (2007) and Zügner et al. (2018), even one and two years after THR, respectively. Beaulieu et al. (2010) and Foucher et al. (2007) speculated that pain-avoidance strategies adopted pre-operatively might persist even after successful surgery, or that alterations are caused by persistent muscle weakness. The one-week intervention period might, therefore, not have been sufficient time for the participants to adopt a normal gait pattern.

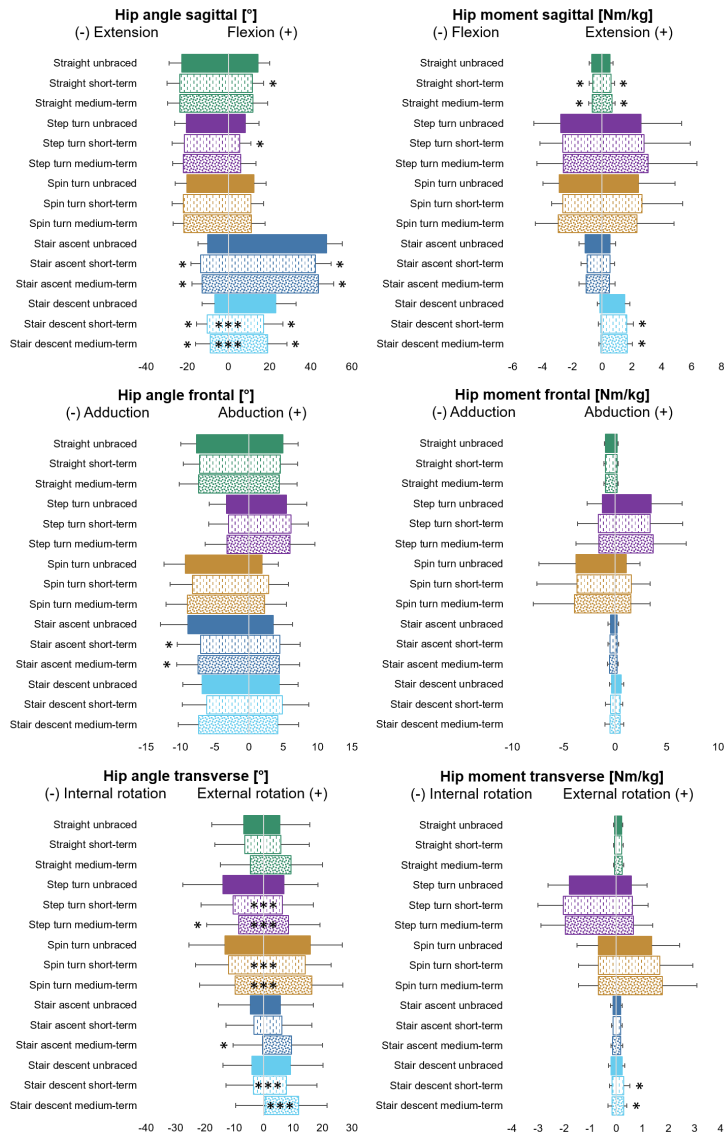


Figure 11.2: Overview of bracing effects on hip joint angles and moments.

Note. Data presented as mean $\pm$ SD. * = significant (p < 0.05) difference between braced and unbraced conditions in peak angles/moments; *** = significant (p < 0.05) difference between braced and unbraced conditions in range of motion.

11.5 Limitations and future research directions

The results of this thesis must be viewed in the context of its limitations, which are as follows:

First of all, both studied groups represent samples of convenience, which raises the risk for bias and thus might limit the generalizability of the findings to the general population of individuals with mild-to-moderate HOA. Moreover, due to ethical considerations, we were unable to obtain radiographic data from the participants of the control group. While control participants did not show symptoms of HOA, subtle changes in joint morphology can not be ruled out. Therefore, future studies should ideally draw random subject samples and rely on functional and radiographic inclusion criteria to establish the absence of illness and disease in control participants (Kraus et al., 2015).

Secondly, we investigated the effects of mild-to-moderate HOA and hip bracing on locomotion biomechanics by analysing kinematics and dynamics of individual joints or bodies. Our results showed that not only is the ipsilateral hip joint affected by HOA, but also the knee joint and the pelvis. However, other joints like the ankle joint, for which HOA-induced alterations in kinematics have been described (Kubota et al., 2007; Ornetti et al., 2011; Schmitt et al., 2015), remained unexamined. Moreover, locomotion tasks are full-body movements, and kinematics of individual joints should be analysed and interpreted in relation to the entire full-body motion. Therefore, future research should include analyses of the whole-body movement, e.g. by application of pattern recognition methods such as principal component analysis or cluster analysis on whole body kinematics, as proposed by Meyer et al. (2015), van Drongelen, Stetter, et al. (2021) and Stetter et al. (2020). This could clarify the interactions of observed HOA or bracing effects, e.g. the coupling of hip, pelvis and lumbar spine motion.

Furthermore, we relied on analysis of joint moments for estimation of joint loading. While joint moments have been shown to correlate with joint loading at the hip (Wesseling et al., 2015) and knee joint (Holder et al., 2020; Manal et al., 2015), they remain surrogate measures. Especially, the role of muscular activity on joint loading is not considered, and observed alterations in joint kinematics cannot be further explained without knowledge of muscular activity. Future studies should include analyses of muscle activity and apply more complex musculoskeletal models to investigate the effects of bracing on muscle activation and joint loading.

Third, bracing in our study caused alterations of joint kinematics. Yet, the mechanism of action of the brace remains unclear, as it is unknown whether alterations stem from mechanical restrictions of the brace or neuromuscular adaptations. Especially limitations in frontal and transverse plane hip kinematics are likely to stem from mechanical restriction of the brace to these movements, as the brace joint only contains one degree of freedom in the sagittal plane.

Including a retention analysis in the form of a motion analysis conducted after bracing interventions, but without applying the brace during measurement, may allow the underlying causes of brace effects to be analysed, as well as how bracing effects are retained over time.

Fourth, the applied JPR test for measurement of hip proprioception demonstrated a minimal detectable change of about 2° and thus, might not be sensitive enough to detect subtle changes in joint proprioception. In our study, the hip flexion movement during the JPR test was performed while standing on one leg. Maintaining balance and simultaneously flexing the hip joint against gravity likely required substantial muscle strength and motor control, potentially masking differences in the sensory quality of hip proprioception. Therefore, more sensitive methods for assessing hip proprioception need to be developed. Performing the JPR test while supine, as done by Alkhamis et al. (2024), might be one way to reduce interference and increase sensitivity.

Lastly, this thesis demonstrated that bracing for one week has an impact on hip pain, functional capacity and locomotion biomechanics. However, the application of braces for treating HOA is more likely to continue for several weeks, months or even years. Especially analyses of brace effects on slowly changing parameters, such as muscle force, require a longer intervention period. Moreover, subtle changes in joint loading of the hip, but also the knee joints, might lower or raise the risk of developing OA in adjacent joints and could impact disease progression at the ipsilateral joint. Therefore, long-term longitudinal studies are warranted to extend and complement the knowledge on short-term and medium-term bracing effects obtained in this thesis.

12 Conclusion

In the past, HOA-induced effects on locomotion biomechanics have been mainly studied during level walking and in individuals with severe forms of HOA.

This thesis demonstrated that individuals with mild-to-moderate HOA show similar limitations in functional capacity and various locomotion biomechanics as individuals with more severe HOA. These limitations include lower functional capacity, lower locomotion velocity, lower peak hip extension, adduction and internal rotation angles, as well as lower hip joint moments in all movement planes. However, as the effect of HOA is less pronounced in this cohort, demanding movement tasks have to be chosen for analysis to uncover these limitations. Moreover, an effect of HOA on postural control during stair walking was found, whereas the assessment of hip proprioception did not yield any differences between individuals with and without HOA.

Subsequently, we demonstrated that one week of bracing with an elastic hip brace can improve hip pain and functional capacity, leading to higher locomotion velocity. Bracing induced changes in sagittal hip and pelvis motion, indicating improved pelvis mobility. Furthermore, brace-induced restriction of peak hip adduction and internal rotation may reduce hip pain and joint loading; however, it could also lead to decreased muscle activity. For some locomotion tasks and planes, bracing caused higher joint moments at the hip joint and partially at the knee joint.

Therefore, the consequences of long-term brace use on muscle force and joint loading should be investigated in the future. Furthermore, the underlying causes of the observed brace-induced changes in locomotion biomechanics need to be further explored, e.g. by analysing retention effects evoked through bracing.

Moreover, the interplay between pelvis and lumbar spine motion and pain remains a topic of controversy. The applied hip brace exerts pressure around the pelvis girdle and onto several trigger points at the SIJ. Future research needs to clarify whether brace-induced increases in pelvic motion are enabled by a stabilisation of the pelvis-spine complex and pain reduction, and whether increased pelvic motion causes greater stress on the lumbar spine, potentially triggering lower back pain with long-term brace use.

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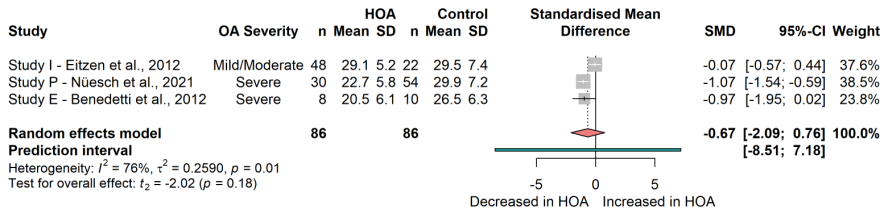
A Appendix

Table A.1: Search terms used for literature search.

Database	Search term	Filters	No. of results
Pubmed	((coxarthr*[Title/Abstract] OR ("degenerative joint"[Title/Abstract] AND hip[Title/Abstract]) OR (hip[Title/Abstract] AND osteoarthr*[Title/Abstract])) AND (kinematic*[Title/Abstract] OR angle*[Title/Abstract] OR "range of motion"[Title/Abstract] OR mobility[Title/Abstract] OR pattern[Title/Abstract] OR goniometric*[Title/Abstract] OR biomechanic*[Title/Abstract]) AND (gait[Title/Abstract] OR walk*[Title/Abstract] OR locomotion[Title/Abstract] OR ambulat*[Title/Abstract] OR stair*[Title/Abstract] OR movement[Title/Abstract]) NOT (fracture[Title] OR perthes[Title] OR amputee[Title] OR rheumat*[Title] OR arthroscopy[Title] OR Arthroplasty[Title]))	Human subjects	637
Web of Science	TS=((coxarthr* OR ("degenerative joint" AND hip) OR (hip AND osteoarthr*)) AND (kinematic* OR angle* OR "range of motion" OR mobility OR pattern OR goniometric* OR biomechanic*) AND (gait OR walk* OR locomotion OR ambulat* OR stair*OR movement)) NOT TI=(fracture OR perthes OR amputee OR rheumat* OR arthroscopy OR Arthroplasty)		1370
Scopus	TITLE-ABS-KEY ((coxarthr* OR ("degenerative joint" AND hip) OR (hip AND osteoarthr*)) AND (kinematic* OR angle* OR "range of motion" OR mobility OR pattern OR goniometric* OR biomechanic*) AND (gait OR walk* OR locomotion OR ambulat* OR stair* OR movement)) AND NOT TITLE (fracture OR perthes OR amputee OR rheumat* OR arthroscopy OR Arthroplasty)		1678

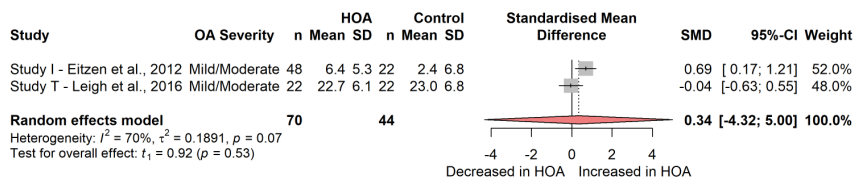
(A)

Gait - Hip Ipsi Sagittal angle at IC



(B)

Gait - Hip Ipsi Sagittal angle at MST



(C)

Gait - Hip Ipsi Sagittal ROM ST

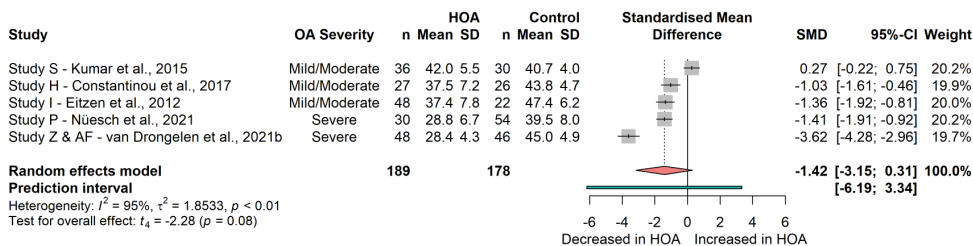
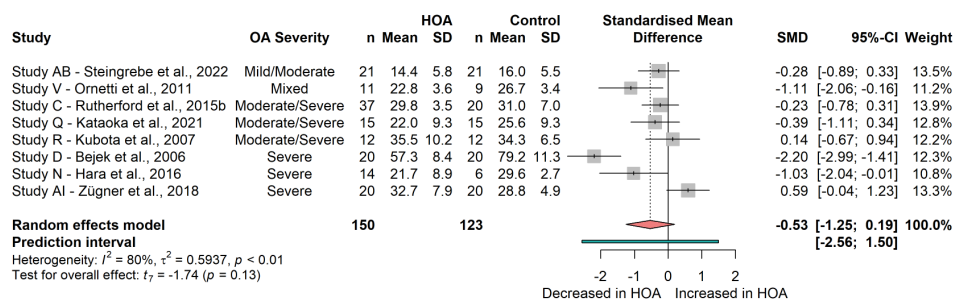


Figure A.1: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) ipsilateral hip sagittal angle at initial contact (IC), (B) ipsilateral hip sagittal angle at mid-stance (MST), (C) ipsilateral hip sagittal range of motion (RoM) across stance phase; during gait.

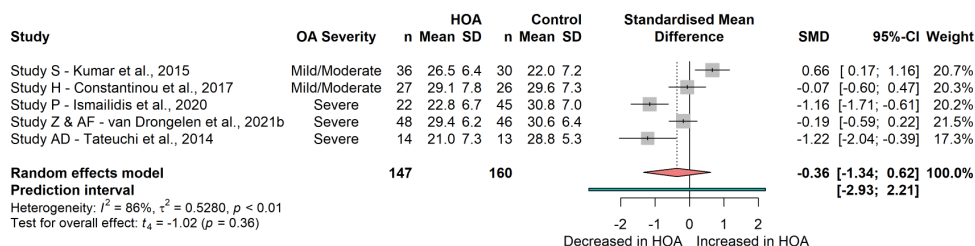
(A)

Gait - Hip Ipsi Peak Flexion GC



(B)

Gait - Hip Ipsi Peak Flexion ST



(C)

Gait - Hip Ipsi Peak Flexion SW

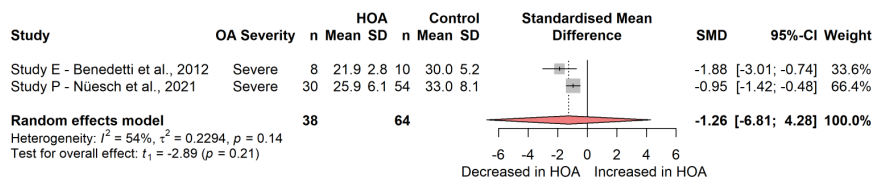
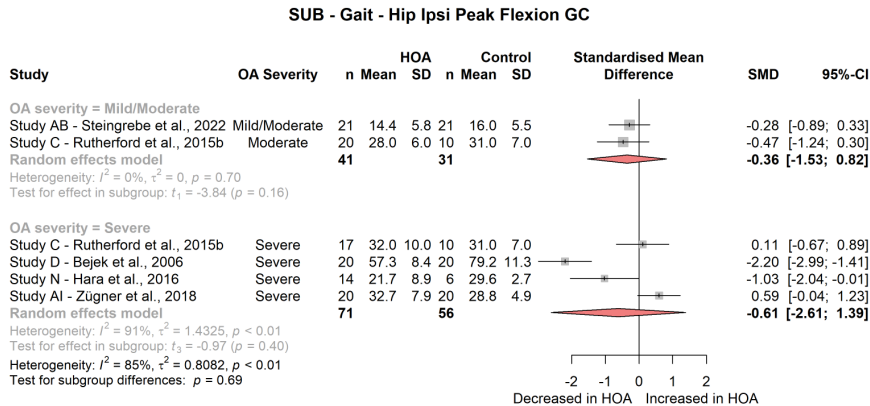


Figure A.2: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) ipsilateral peak hip flexion during gait cycle (GC), (B) ipsilateral peak hip flexion during stance phase (ST), (C) ipsilateral peak hip flexion during swing phase (SW); during gait.

(A)



(B)

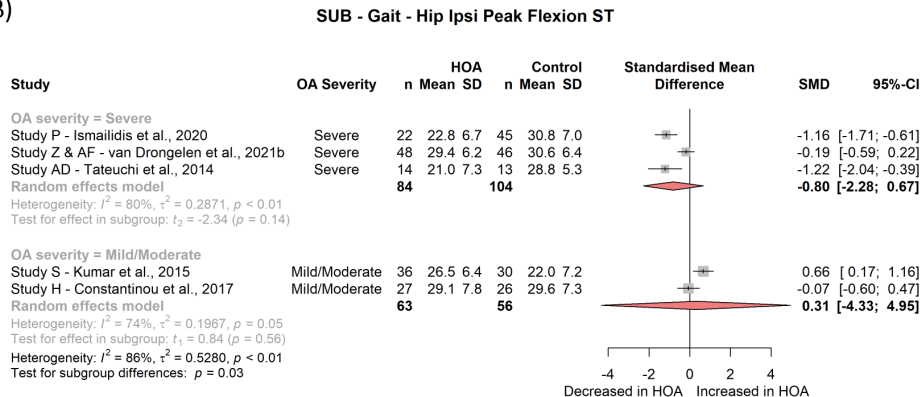
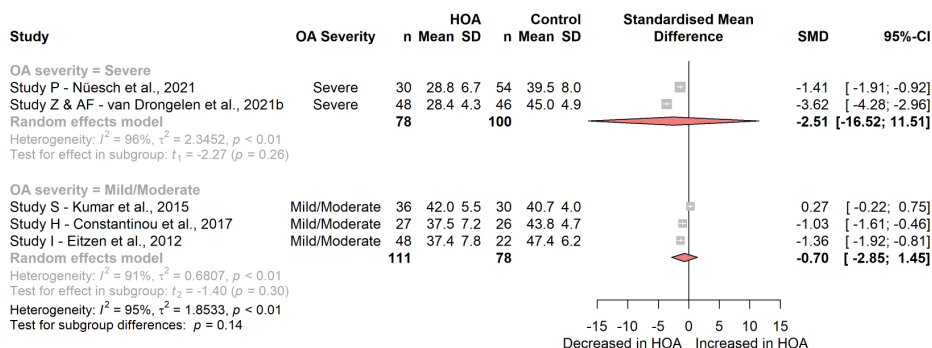


Figure A.3: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for subgroup analyses on (A) ipsilateral peak hip flexion during gait cycle (GC), (B) ipsilateral peak hip flexion during stance phase (ST); during gait.

(A)

SUB - Gait - Hip Ipsi Sagittal ROM ST



(B)

SUB - Gait - Hip Ipsi Transverse ROM GC

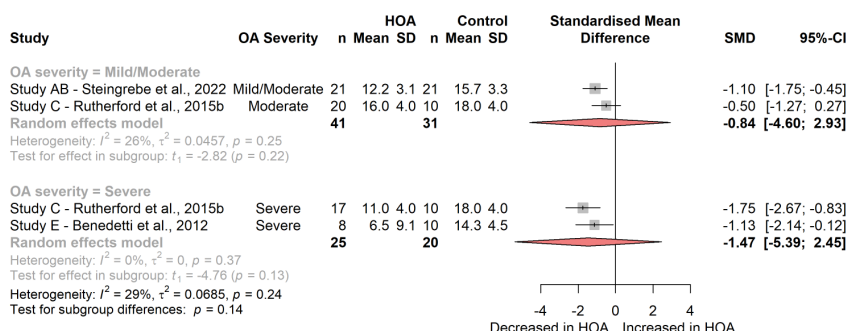


Figure A.4: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for subgroup analyses on (A) ipsilateral hip sagittal range of motion (RoM) across stance phase (ST), (B) ipsilateral hip transverse RoM across gait cycle (GC); during gait

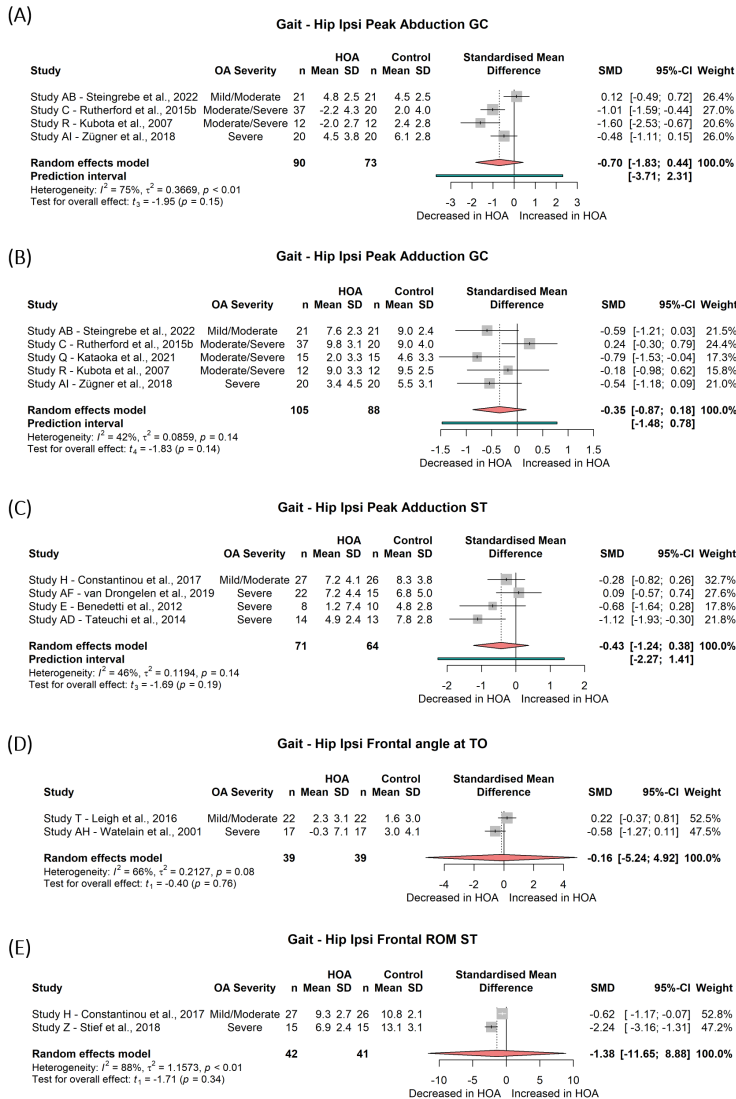
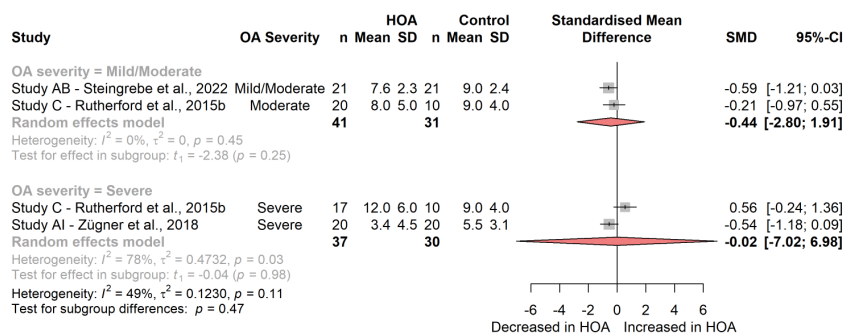


Figure A.5: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) ipsilateral peak hip abduction during gait cycle (GC), (B) ipsilateral peak hip adduction during GC, (C) ipsilateral peak hip adduction during stance phase (ST), (D) ipsilateral frontal hip angle at toe-off (TO), (E) ipsilateral hip frontal range of motion (RoM) across ST; during gait.

(A)

SUB - Gait - Hip Ipsi Peak Adduction GC



(B)

SUB - Gait - Hip Ipsi Peak Abduction GC

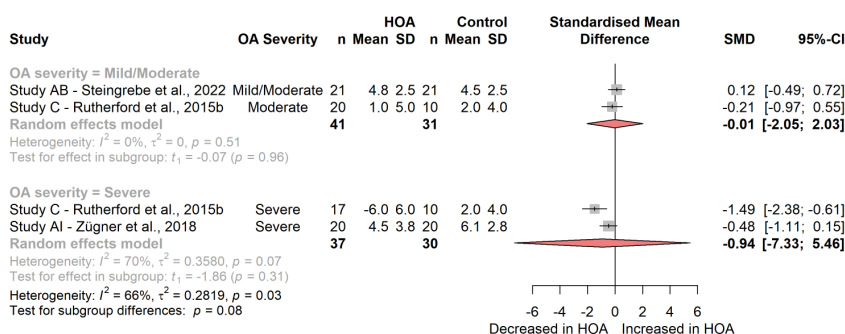
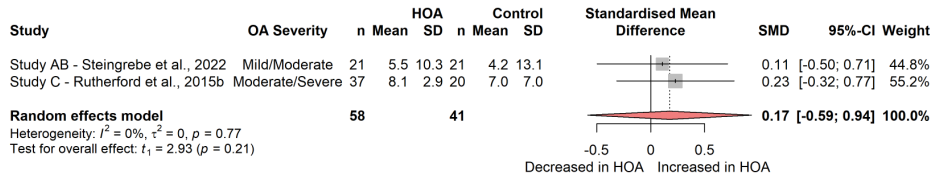


Figure A.6: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for subgroup analyses on (A) ipsilateral hip peak adduction during gait cycle (GC), (B) ipsilateral hip peak abduction during GC; during gait

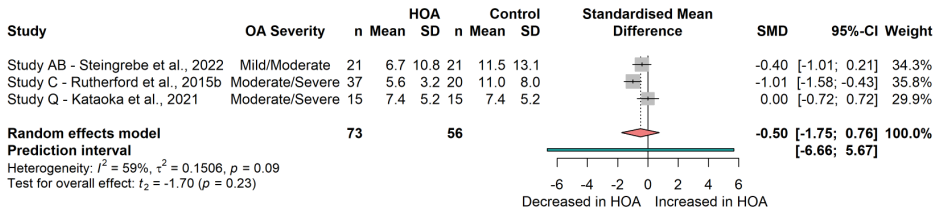
(A)

Gait - Hip Ipsi Peak External Rotation GC



(B)

Gait - Hip Ipsi Peak Internal Rotation GC



(C)

Gait - Hip Ipsi Transverse angle at TO

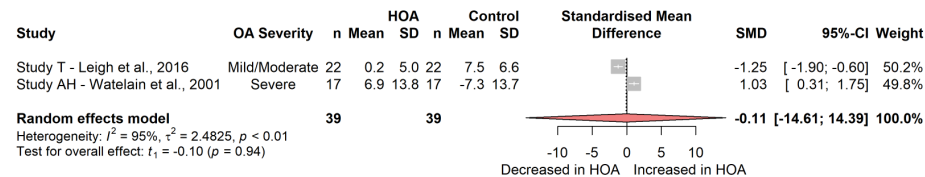


Figure A.7: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) ipsilateral hip peak external rotation during gait cycle (GC), (B) ipsilateral hip peak internal rotation during GC, (C) ipsilateral hip transverse angle at toe-off (TO); during gait.

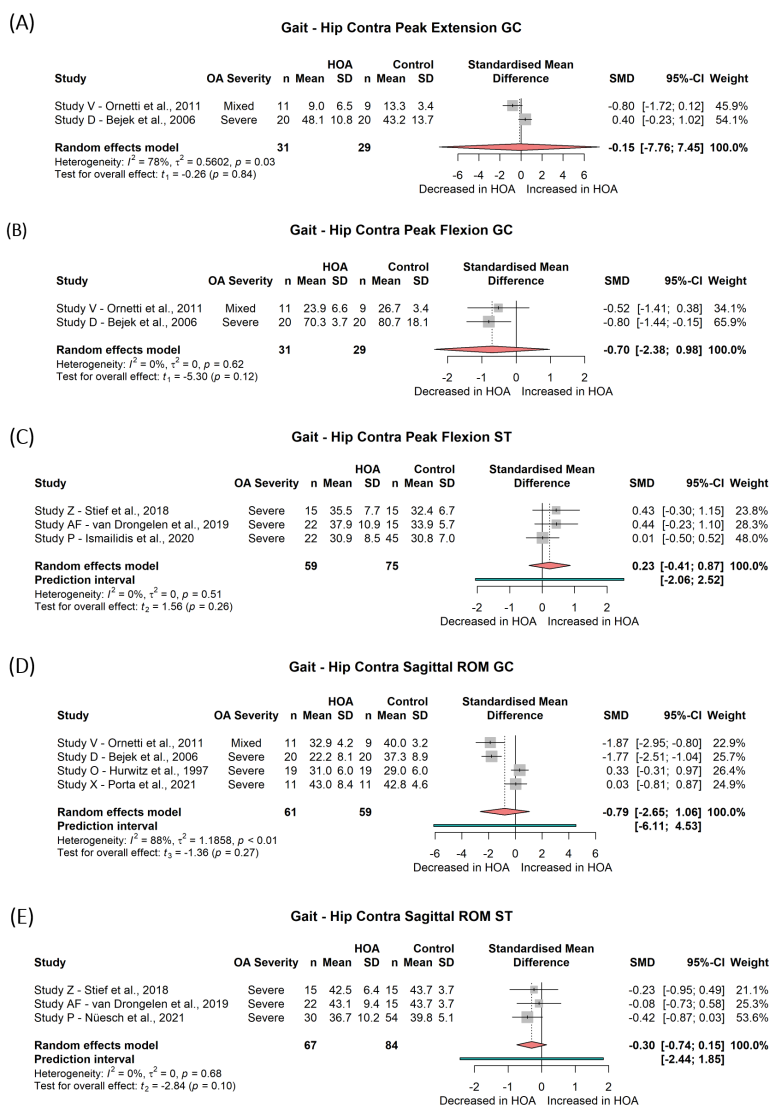
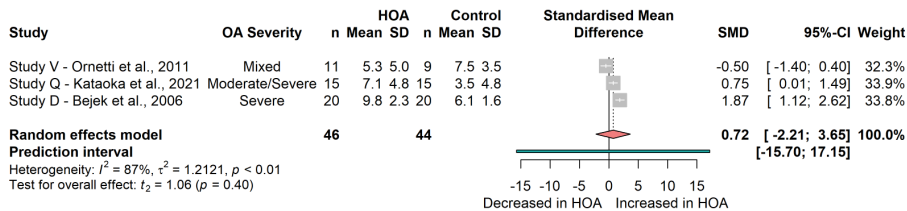


Figure A.8: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) contralateral peak hip extension during gait cycle (GC), (B) contralateral peak hip flexion during GC, (C) contralateral peak hip flexion during stance phase (ST), (D) contralateral hip sagittal range of motion (RoM) across GC, (E) contralateral hip sagittal RoM ST; during gait.

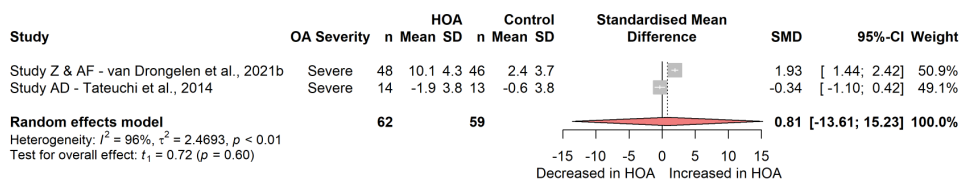
(A)

Gait - Knee Ipsi Peak Extension GC



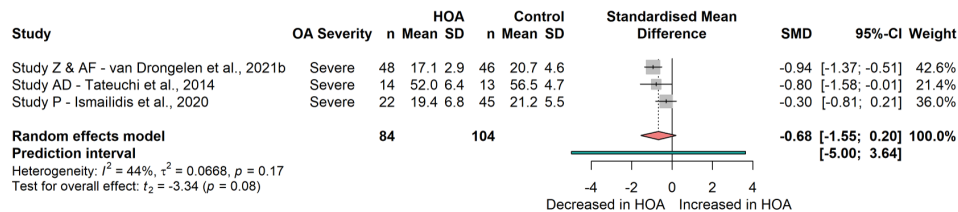
(B)

Gait - Knee Ipsi Peak Extension ST



(C)

Gait - Knee Ipsi Peak Flexion ST



(D)

Gait - Knee Ipsi Sagittal ROM ST

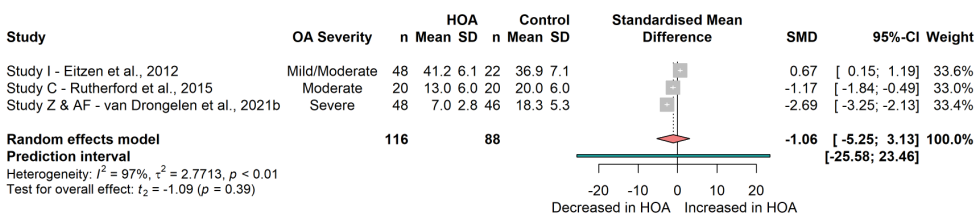
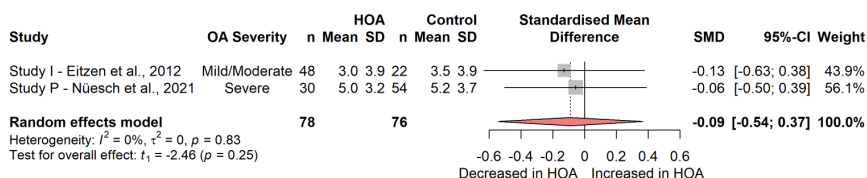


Figure A.9: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) ipsilateral peak knee extension during gait cycle (GC), (B) ipsilateral peak knee extension during stance phase (ST), (C) ipsilateral peak knee flexion ST, (D) ipsilateral knee sagittal range of motion (RoM) across ST; during gait.

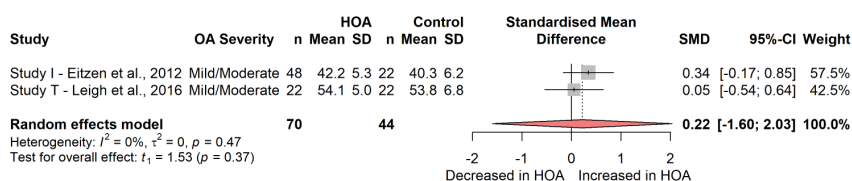
(A)

Gait - Knee Ipsi Sagittal angle at IC



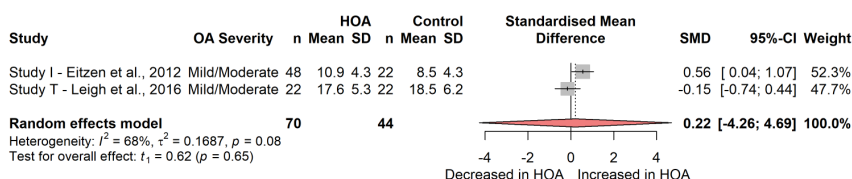
(B)

Gait - Knee Ipsi Sagittal angle at TO



(C)

Gait - Knee Ipsi Sagittal angle at MSt



(D)

Gait - Knee Ipsi Sagittal angle at Peak hip extension

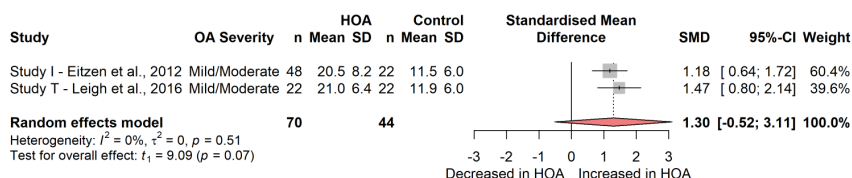


Figure A.10: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) ipsilateral knee sagittal angle at initial contact (IC), (B) ipsilateral knee sagittal angle at toe-off (TO), (C) ipsilateral knee sagittal angle at midstance (MSt), (D) ipsilateral knee sagittal angle at peak hip extension; during gait.

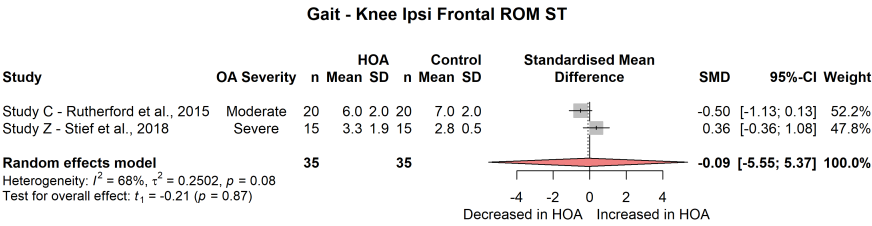
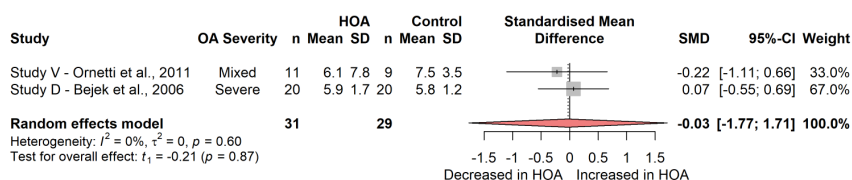


Figure A.11: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for ipsilateral frontal knee range of motion (RoM) across stance phase (ST) during gait.

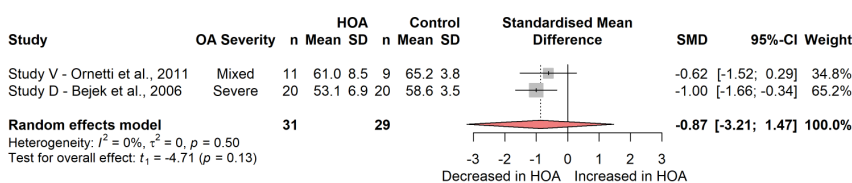
(A)

Gait - Knee Contra Peak Extension GC



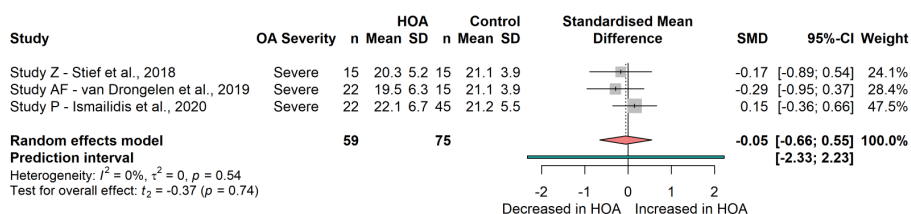
(B)

Gait - Knee Contra Peak Flexion GC



(C)

Gait - Knee Contra Peak Flexion ST



(D)

Gait - Knee Contra Frontal ROM ST

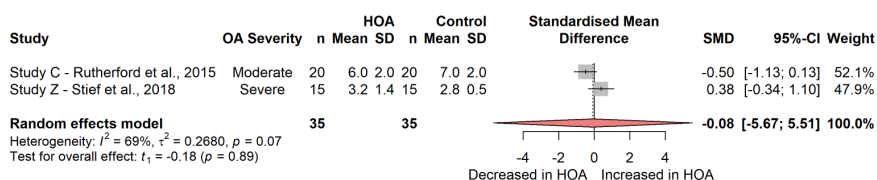


Figure A.12: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) contralateral peak knee extension during gait cycle (GC), (B) contralateral peak knee flexion during GC, (C) contralateral peak knee flexion during stance phase (ST), (D) contralateral knee frontal range of motion (RoM) across ST; during gait.

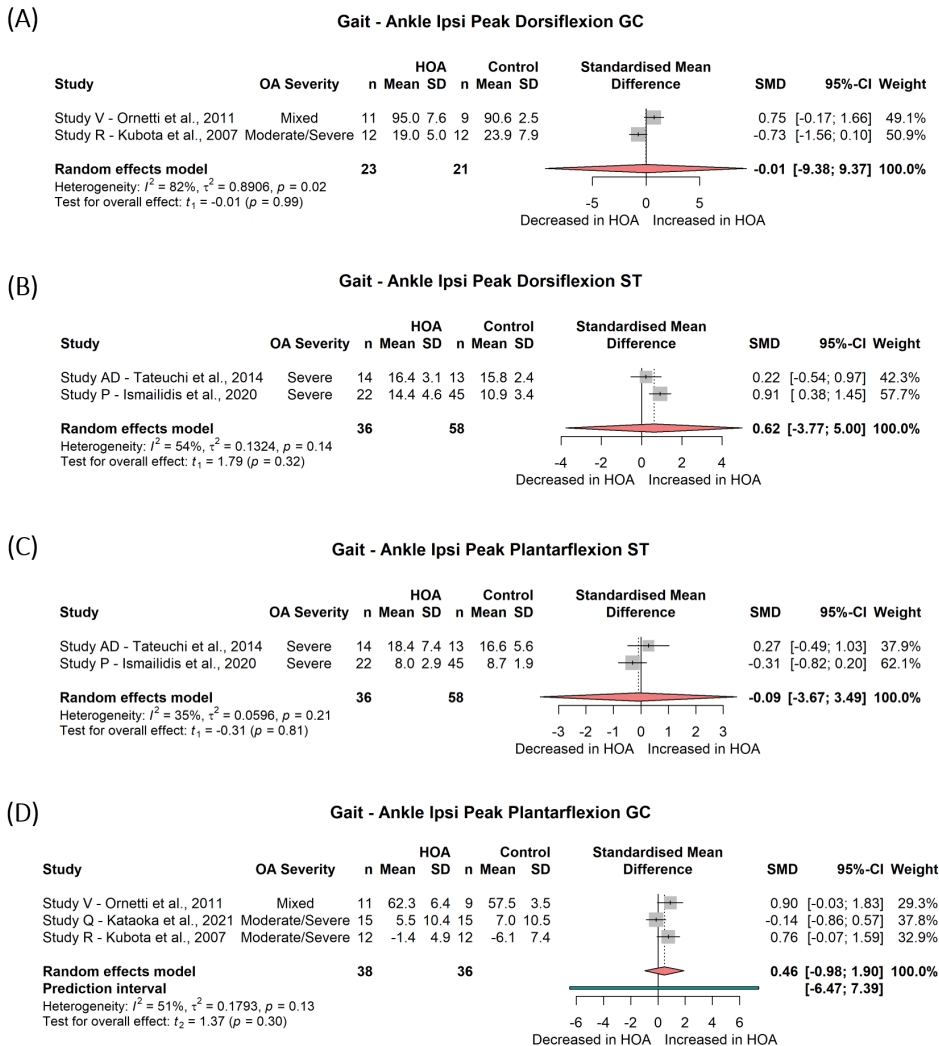


Figure A.13: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) ipsilateral peak ankle dorsiflexion during gait cycle (GC), (B) ipsilateral peak ankle dorsiflexion during stance phase (ST), (C) ipsilateral peak ankle plantarflexion ST, (D) ipsilateral peak ankle plantarflexion GC; during gait.

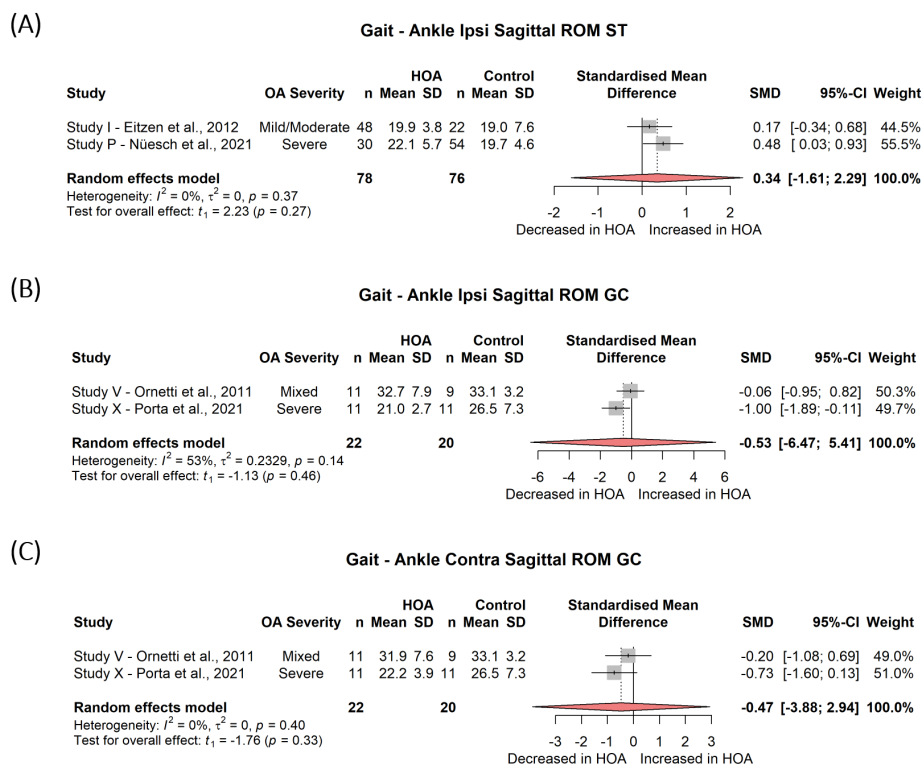


Figure A.14: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) ipsilateral ankle sagittal range of motion (RoM) across stance phase (ST), (B) ipsilateral ankle sagittal RoM across gait cycle (GC), (C) contralateral ankle sagittal RoM across GC; during gait.

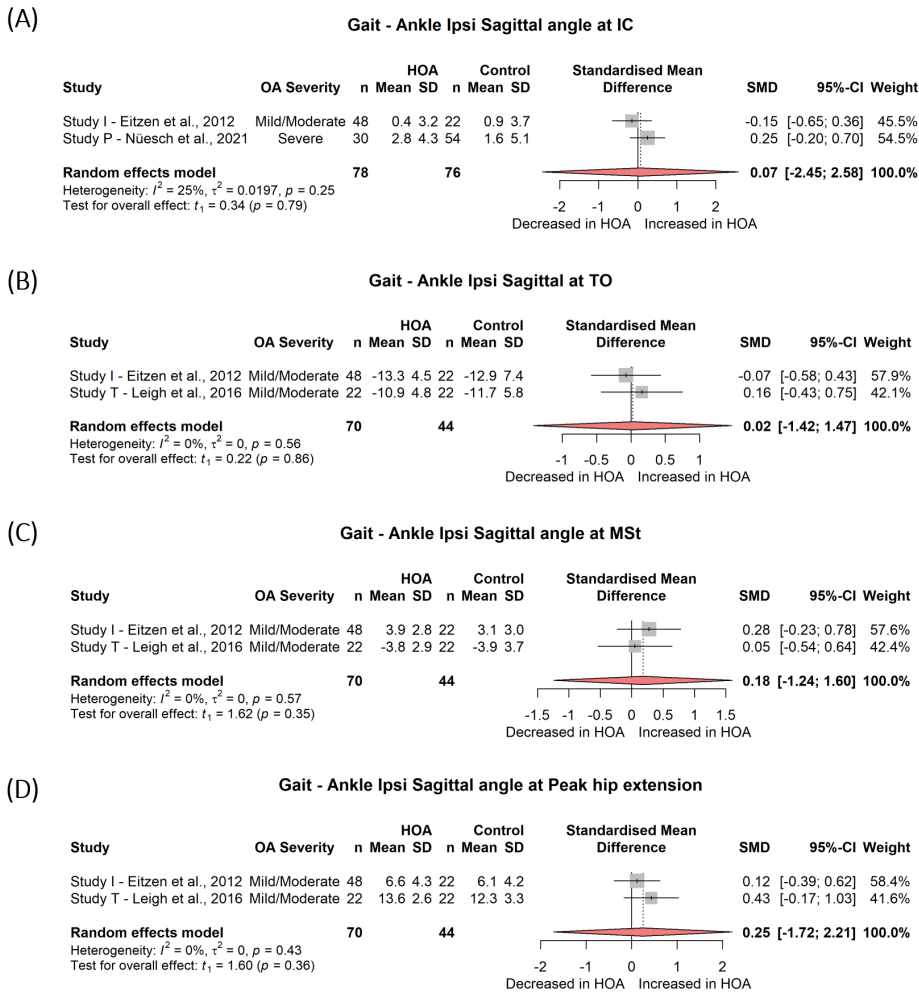


Figure A.15: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) ipsilateral ankle sagittal angle at initial contact (IC), (B) ipsilateral ankle sagittal angle at toe-off (TO), (C) ipsilateral ankle sagittal angle at midstance (MSt), (D) ipsilateral ankle sagittal angle at peak hip extension; during gait.

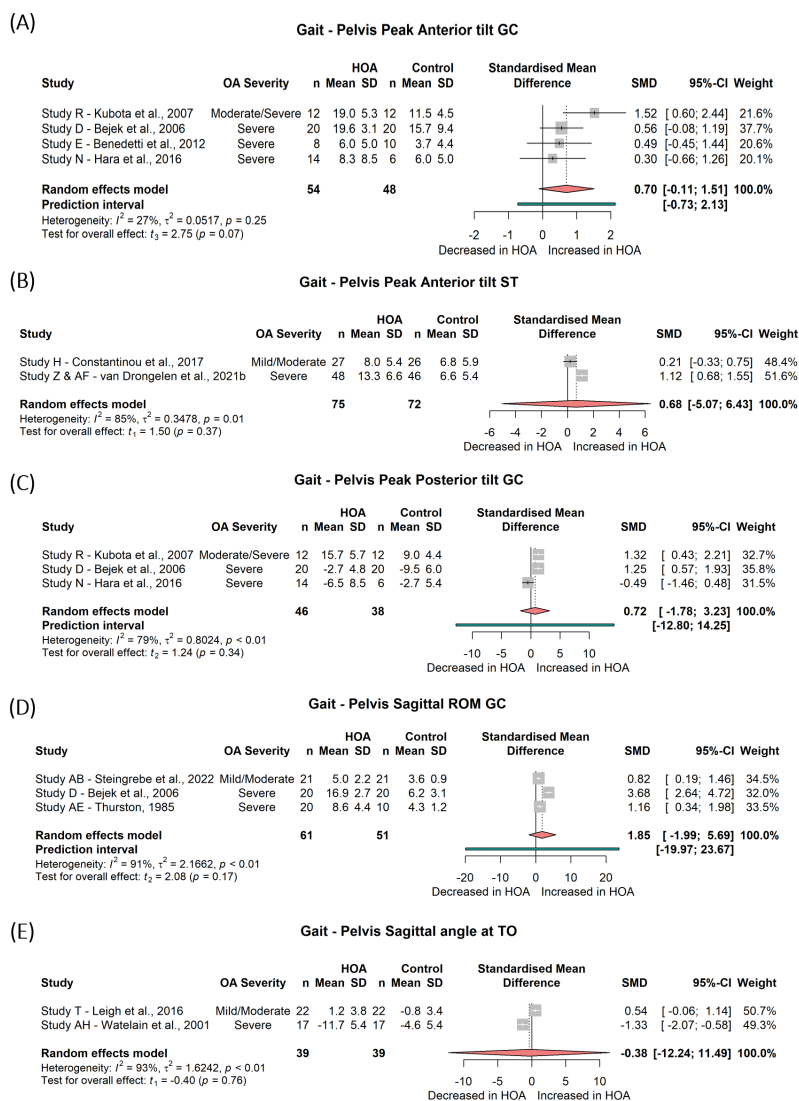
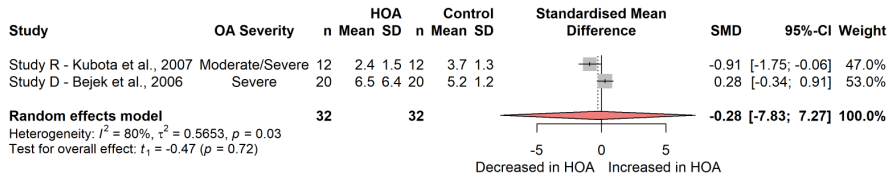


Figure A.16: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) pelvis peak anterior tilt during gait cycle (GC), (B) pelvis peak anterior tilt during stance phase (ST), (C) pelvis peak posterior tilt during GC, (D) pelvis sagittal range of motion (RoM) across GC, (E) pelvis sagittal angle at toe-off (TO); during gait.

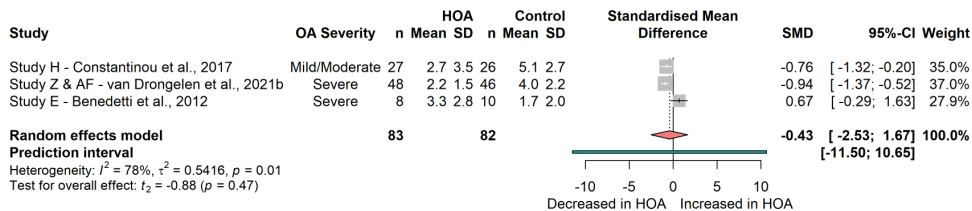
(A)

Gait - Pelvis Peak Superior Obliquity GC



(B)

Gait - Pelvis Peak Superior Obliquity ST



(C)

Gait - Pelvis Peak Inferior Obliquity GC

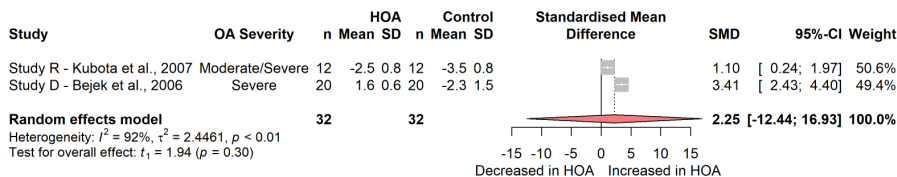
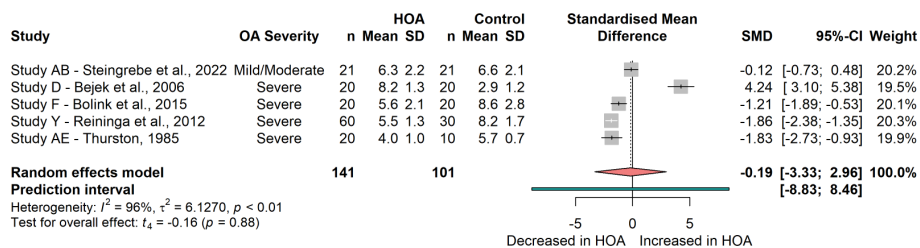


Figure A.17: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) pelvis peak superior obliquity during gait cycle (GC), (B) pelvis peak superior obliquity during stance phase (ST), (C) pelvis peak inferior obliquity during GC; during gait.

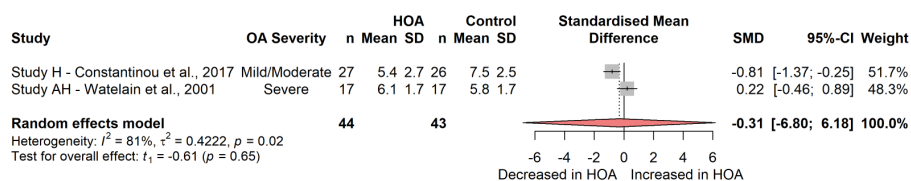
(A)

Gait - Pelvis Frontal ROM GC



(B)

Gait - Pelvis Frontal ROM ST



(C)

Gait - Pelvis Frontal angle at TO

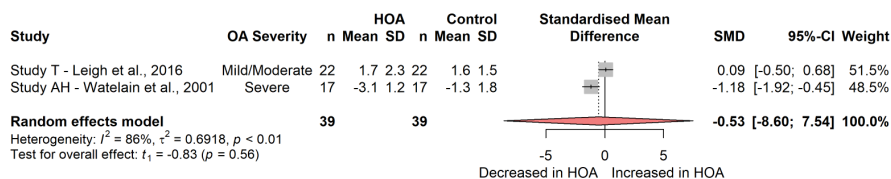


Figure A.18: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) pelvis frontal range of motion (RoM) across gait cycle (GC), (B) pelvis frontal RoM across stance phase (ST), (C) pelvis frontal angle at toe-off (TO); during gait.

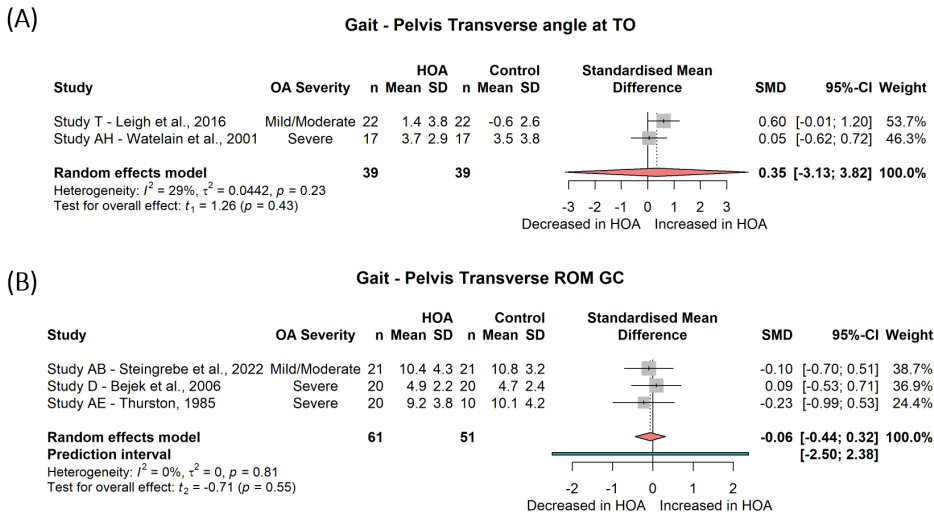


Figure A.19: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) pelvis transverse angle at toe-off (TO), (B) pelvis transverse range of motion (RoM) across gait cycle (GC); during gait.

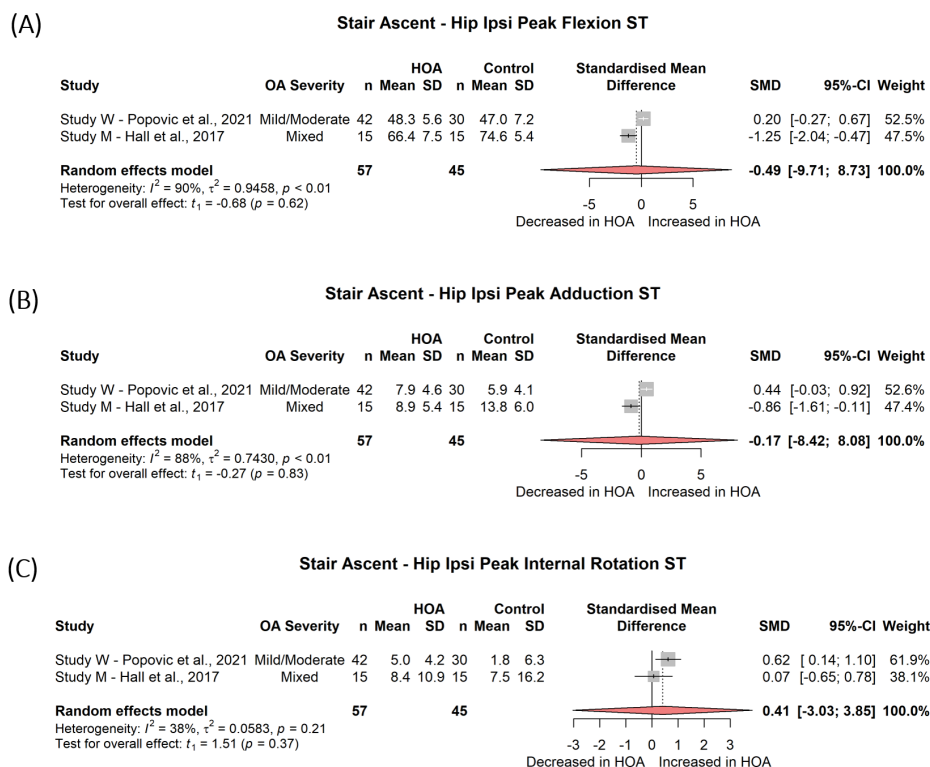


Figure A.20: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) ipsilateral peak hip flexion during stance phase (ST), (B) ipsilateral peak hip adduction during ST, (C) ipsilateral peak hip internal rotation during ST; during stair ascent.

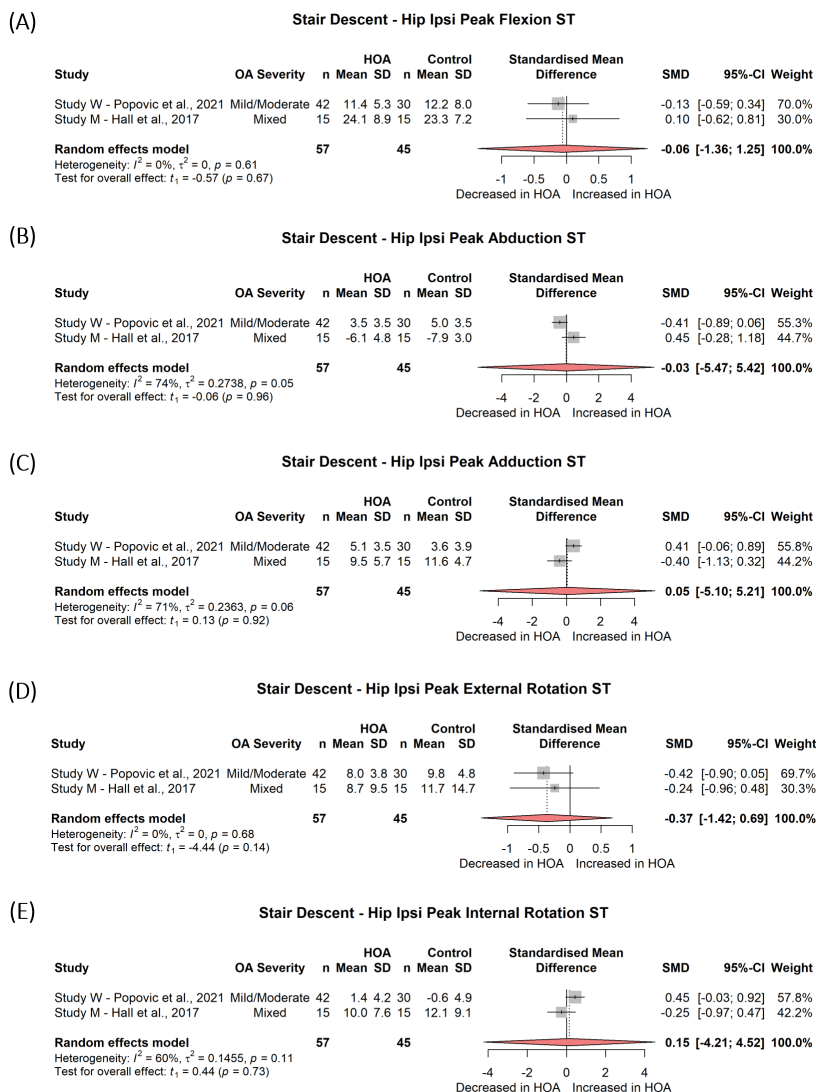


Figure A.21: Forest plot of standardised and pooled effect sizes (random-effects model) with I^2 heterogeneity statistics for (A) ipsilateral peak hip flexion during stance phase (ST), (B) ipsilateral peak hip abduction during ST, (C) ipsilateral peak hip adduction during ST, (D) ipsilateral peak hip external rotation during ST, (E) ipsilateral peak hip internal rotation during ST; during stair descent.

Table A.2: P-values of Shapiro-Wilk tests, Levene tests and mixed-model ANOVA.

Parameter	Shapiro-Wilks Test (p)						Levene-Test (p)	Mixed-Model ANOVA (p)			
	Healthy			HOA				Group	Main effects		Interaction effect
	Straight	Step Turn	Spin Turn	Straight	Step Turn	Spin Turn			Group	Movement	
CoM velocity IC [m/s]	0.125	0.379	0.995	0.906	0.333	0.241	0.497	0.187	<0.001	0.333	
CoM velocity TO [m/s]	0.029	0.363	0.654	0.775	0.139	0.322	0.837	0.123	<0.001	0.139	
Stance phase duration [s]	0.710	0.036	0.666	0.348	0.271	0.003	0.198	0.292	<0.001	0.271	
Peak hip flexion [°]	0.768	0.345	0.910	0.269	0.206	0.942	0.564	0.464	<0.001	0.206	
Peak hip extension [°]	0.562	0.063	0.532	0.069	0.141	0.897	0.311	0.002	<0.001	0.141	
Sagittal hip RoM [°]	0.242	0.794	0.613	0.332	0.041	0.364	0.128	0.001	<0.001	0.041	
Peak hip abduction [°]	0.531	0.147	0.886	0.258	0.001	0.282	0.425	0.997	<0.001	0.001	
Peak hip adduction [°]	0.353	0.051	0.513	0.320	<0.001	0.988	0.206	0.215	<0.001	<0.001	
Frontal hip RoM [°]	0.369	0.100	0.163	0.059	0.081	0.088	0.317	0.154	<0.001	0.081	
Peak hip ext. rotation [°]	0.592	0.479	0.969	0.097	0.281	0.135	0.089	0.849	<0.001	0.281	
Peak hip int. rotation [°]	0.896	0.939	0.948	0.042	0.046	0.314	0.244	0.099	<0.001	0.046	
Transverse hip RoM [°]	0.632	0.560	0.122	0.703	0.006	0.256	0.379	<0.001	<0.001	0.006	

Table A.2: P-values of Shapiro-Wilk tests, Levene tests and mixed-model ANOVA.

Parameter	Shapiro-Wilks Test (p)						Levene-Test (p)	Mixed-Model ANOVA (p)			
	Healthy			HOA				Group	Main effects		Interaction effect
	Straight	Step Turn	Spin Turn	Straight	Step Turn	Spin Turn			Group	Movement	
Peak hip extension moment [Nm/kg]	<0.001	0.002	0.001	0.181	0.851	0.001	0.852	0.349	0.006	0.851	
Peak hip flexion moment [Nm/kg]	<0.001	0.395	0.537	0.653	0.493	0.423	0.132	0.823	<0.001	0.493	
Peak hip abduction moment [Nm/kg]	0.149	0.003	0.002	0.048	0.740	0.001	0.616	0.637	<0.001	0.740	
Peak hip adduction moment [Nm/kg]	0.002	0.001	0.001	0.088	0.730	0.002	0.595	0.667	<0.001	0.730	
Peak hip ext. rotation moment [Nm/kg]	0.859	0.012	0.017	0.092	0.175	0.039	0.634	0.554	<0.001	0.175	
Peak hip int. rotation moment [Nm/kg]	<0.001	0.839	<0.001	0.244	0.198	0.003	0.322	0.319	<0.001	0.198	

Note: HOA = hip osteoarthritis group; IC = initial contact; TO = toe off; CoM = centre of mass; RoM = range of motion.

Table A.3: Bonferroni corrected p-values from pairwise comparisons of significant ANOVA main effects for movement task using dependent sample t-tests.

Parameter	Bonferroni corrected dependent sample t-tests (p)		
	Straight vs. Step Turn	Straight vs. Spin Turn	Step Turn vs. Spin Turn
CoM velocity IC [m/s]	<0.001	<0.001	0.364
CoM velocity TO [m/s]	<0.001	<0.001	1.000
Stance phase duration [s]	<0.001	<0.001	<0.001
Peak hip flexion [°]	<0.001	0.001	<0.001
Peak hip extension [°]	<0.001	<0.001	1.000
Sagittal hip RoM [°]	<0.001	<0.001	<0.001
Peak hip abduction [°]	<0.001	<0.001	<0.001
Peak hip adduction [°]	<0.001	<0.001	<0.001
Frontal hip RoM [°]	<0.001	1.000	<0.001
Peak hip ext. rotation [°]	<0.001	<0.001	<0.001
Peak hip int. rotation [°]	<0.001	<0.001	0.162
Transverse hip RoM [°]	<0.001	<0.001	<0.001
Peak hip extension moment [Nm/kg]	<0.001	<0.001	1.000
Peak hip flexion moment [Nm/kg]	<0.001	<0.001	1.000
Peak hip abduction moment [Nm/kg]	<0.001	<0.001	<0.001
Peak hip adduction moment [Nm/kg]	0.910	<0.001	<0.001
Peak hip ext. rotation moment [Nm/kg]	0.004	<0.001	<0.001
Peak hip int. rotation moment [Nm/kg]	<0.001	<0.001	<0.001

Note. IC = initial contact; TO = toe off; CoM = centre of mass; RoM = range of motion.

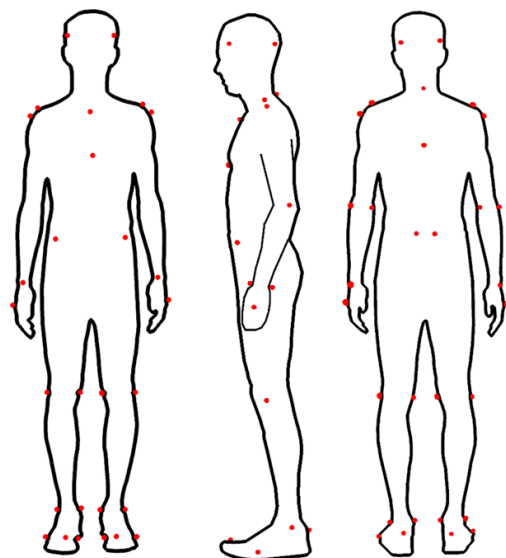


Figure A.22: Locations of the 42 retroreflective markers from frontal, side and rear view.

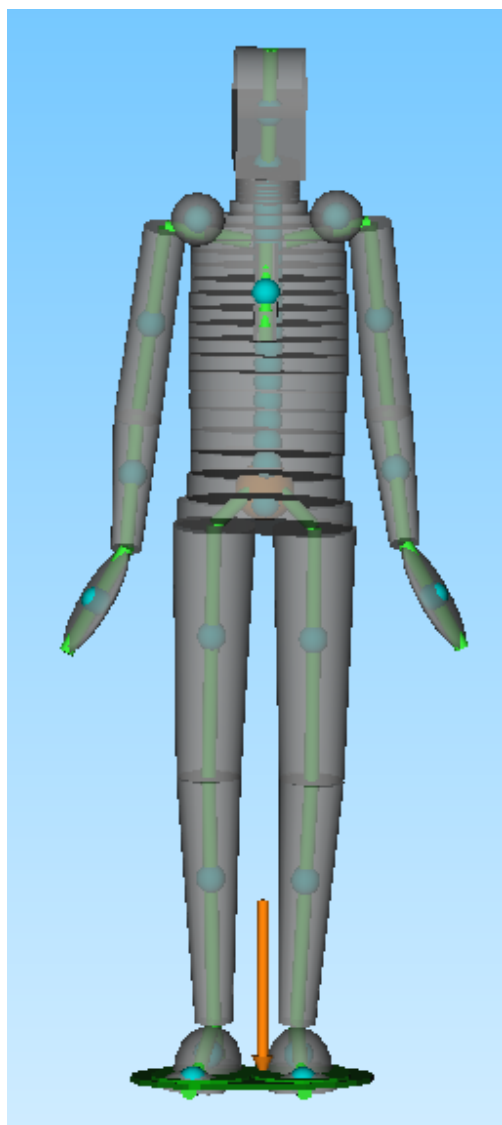


Figure A.23: Geometrical shapes of the model segments.

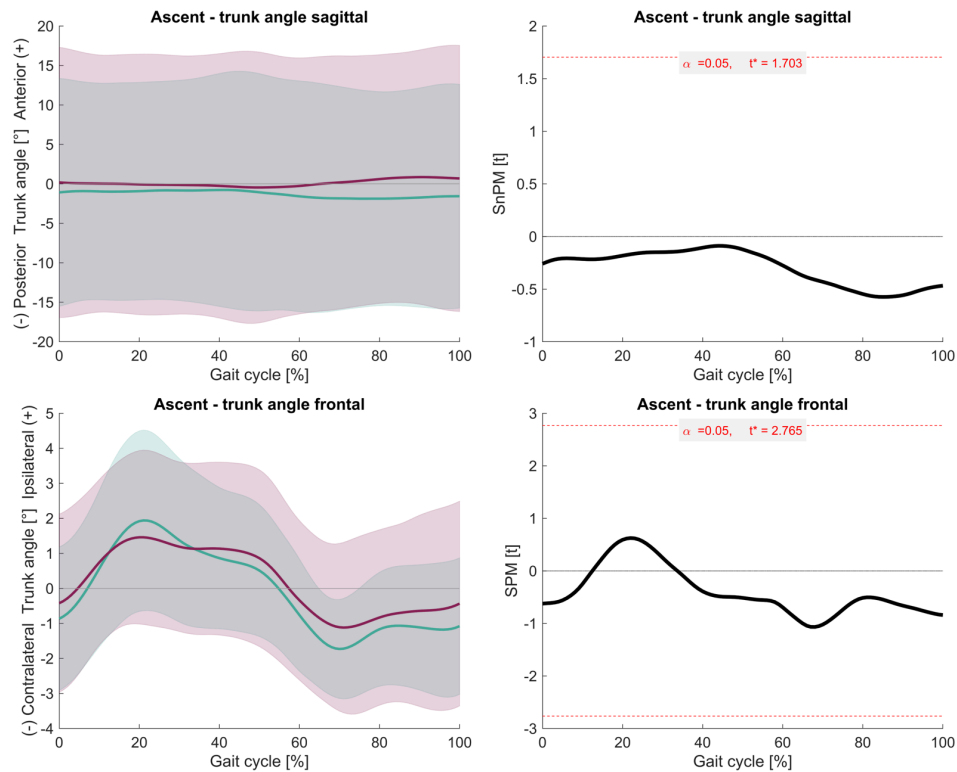


Figure A.24: Trunk angle time curves during stair ascent with results from SPM and SnPM analyses.
Note. Data presented as mean $\pm$ SD. Green line = hip osteoarthritis group; purple line = control group.

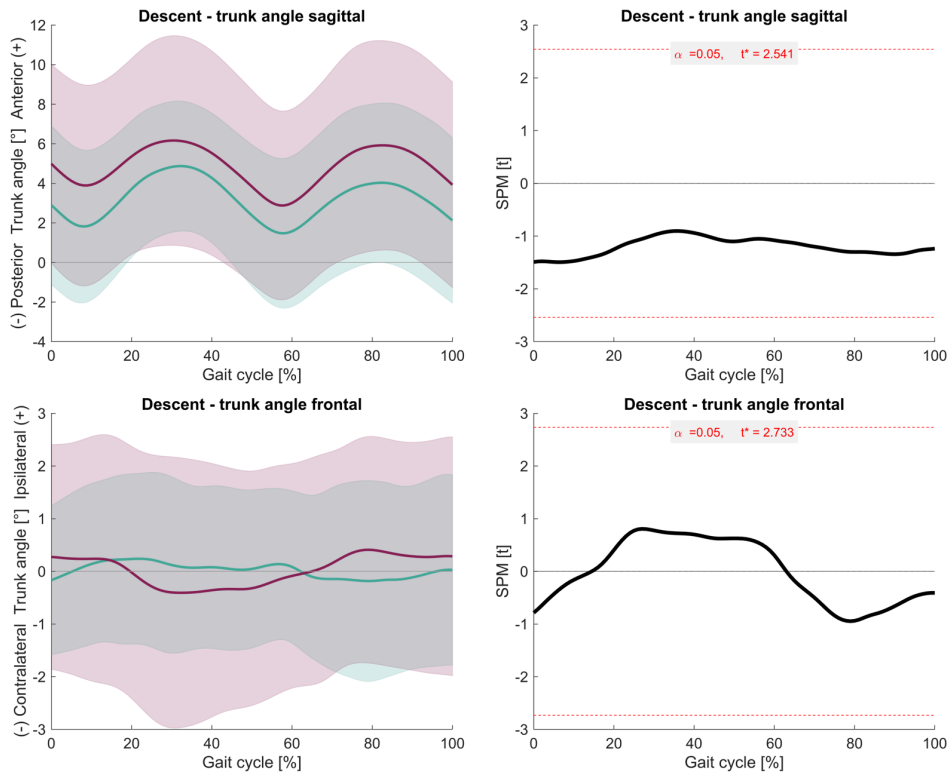


Figure A.25: Trunk angle time curves during stair descent with results from SPM analysis.
Note. Data presented as mean $\pm$ SD. Green line = hip osteoarthritis group; purple line = control group.

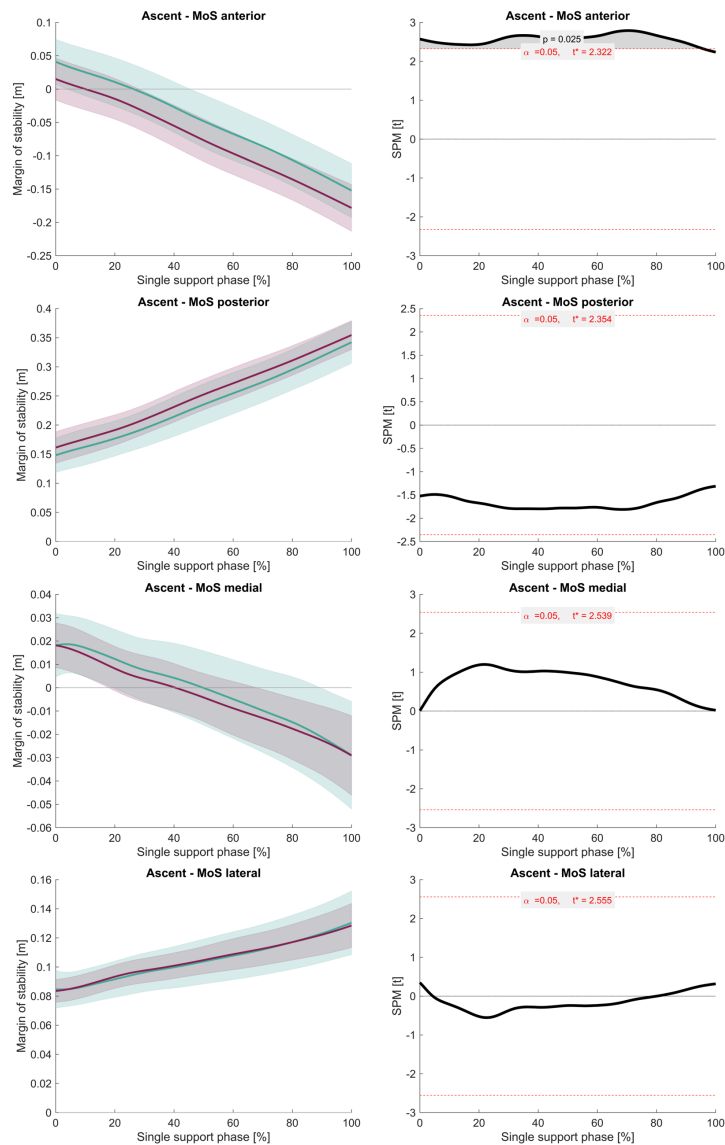


Figure A.26: Margin of stability (MoS) in anterior, posterior, medial and lateral directions during stair ascent with results from SPM analysis.

Note. Data presented as mean $\pm$ SD. Green line = hip osteoarthritis group; purple line = control group.

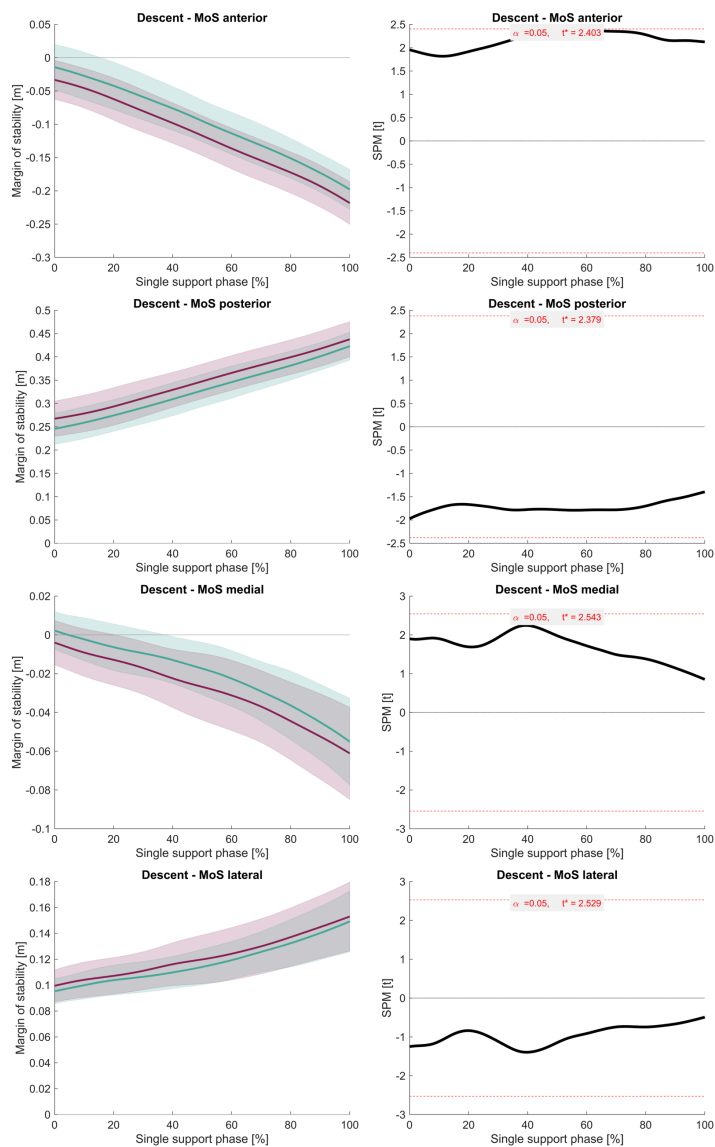


Figure A.27: Margin of stability (MoS) in anterior, posterior, medial and lateral directions during stair descent with results from SPM analysis.

Note. Data presented as mean $\pm$ SD. Green line = hip osteoarthritis group; purple line = control group.

Table A.4: Definitions of the locations of the 42 retroreflective markers.

Segment (n of markers)	Marker name	Description
Head (4)	LFHD/RFHD	Left & right forehead between os frontal and os parietal
	LBHD/RBHD	In a horizontal line to the right/left forehead marker
Shoulder (6)	C7	Spinous process of the 7 th cervical vertebra
	LACR/RACR	Left & right acromio-clavicular joint
	LHUM/RHUM	Left & right humeral head
	CLAV	Centred between the articuli sterno-clavicularis
Thorax (2)	STRN	Xiphoid process of the sternum
	T10	Spinous process of the 10 th thoracic vertebrae
Hip (4)	LASI/RASI	Left & right anterior superior iliac spine
	LPSI/RPSI	Left & right posterior superior iliac spine
Upper arm (4)	LELBlat/RELBlat	Left & right lateral epicondyle of the humerus
	LELBmed/RELBmed	Left & right medial epicondyle of the humerus
Forearm (4)	LWRlIat/RWRlIat	Left & right ulna-styloid process
	LWRlmed/RWRlmed	Left & right radius-styloid process
Hand (2)	LFIN/RFIN	Dorsal side of the proximal third metacarpal bone
Leg (4)	LKNElat/RKNElat	Left & right epicondylus lateralis femoris
	LKNEmed/RKNEmed	Left & right epicondylus medialis femoris
Ankle (6)	LMALlat/RMALlat	Fibula apex of the lateral malleolus
	LMALmed/RMALmed	Tibia apex of the medial malleolus
	LHEEL/RHEEL	Posterior surface of the calcaneus
Foot (6)	LFOOTlat/LFOOTmed	First metatarsophalangeal joint
	LFOOTmed/RFOOTmed	Fifth metatarsophalangeal joint
	LTOE/RTOE	1 st phalanx distalis

Table A.5: Anthropometric measures taken for individual model scaling.

Parameters measured manually	Parameters extracted from marker data
Height	Foot height (left/right)
Weight	Knee joint height (left/right)
Functional leg length (left/right)	Ankle joint height (left/right)
Waist height	Height 7 th cervical vertebra
Xiphoid process height	Foot width (left/right)
Atlas height	Shoulder width
Foot length (left/right)	Upper arm length (left/right)
Largest calf circumference (left/right)	Lower arm length (left/right)
Largest thigh circumference (left/right)	Knee width (left/right)
Smallest lower leg circumference (left/right)	Ankle width (left/right)
Pelvis width	Pate width
Waist width	Pate depth
Largest pelvis circumference	Suprasternal notch height
Waist circumference	Calcaneus height (left/right)
Thorax width	Midfoot length (left/right)
Thorax depth	5 th metatarsal height (left/right)
Largest upper arm circumference (left/right)	Toe height (left/right)
Largest lower arm circumference (left/right)	Elbow width (left/right)
Smallest lower arm circumference (left/right)	Wrist width (left/right)
Hand length (left/right)	Sternum length
Neck circumference	
Pate length	

Table A.6: Results of bracing effects on discrete kinematic and kinetic parameters of the knee joint during the step turn.

Parameter during step turn		Base-line	Short-term	Medium-term	ANOVA/ Friedman	B vs. S	B vs. M	S vs. M
		Mean (SD)	Mean (SD)	Mean (SD)	p (η_p^2)	MD (95% CI)	MD (95% CI)	MD (95% CI)
Sagittal plane	Knee peak extension[°]	11.05 (4.17)	9.16 (4.13)	8.85 (4.72)	0.10 (0.12)	1.89 (0.74 3.04)	2.20 (-0.35 4.76)	0.31 (-2.07 2.70)
	Knee peak flexion [°]	44.37 (4.74)	44.19 (4.72)	42.62 (6.76)	0.26 (0.06)	0.17 (-1.55 1.90)	1.75 (-0.78 4.27)	1.57 (-1.30 4.45)
	Knee sagittal RoM [°]	33.32 (5.25)	35.03 (5.09)	33.77 (5.86)	0.19 (0.08)	-1.71 (-3.20 -0.23)	-0.45 (-2.83 1.93)	1.26 (-0.62 3.14)
	Knee peak extension moment [Nm/kg]	-2.37 (1.59)	-1.88 (1.22)	-1.88 (1.42)	0.08 (0.26)	-0.50 (-1.24 0.25)	-0.50 (-1.16 0.16)	0.00 (-0.52 0.52)
	Knee peak flexion moment [Nm/kg]	0.72 (1.22)	1.08 (1.49)	1.44 (1.95)	0.65 (0.17)	-0.36 (-0.90 0.18)	-0.72 (-1.36 -0.08)	-0.36 (-0.74 0.02)
Frontal plane	Knee peak adduction [°]	-2.66 (3.66)	-2.53 (4.93)	-1.63 (4.11)	0.42 (0.04)	-0.14 (-1.95 1.68)	-1.03 (-2.50 0.44)	-0.89 (-2.03 1.03)
	Knee peak abduction [°]	2.36 (4.18)	2.37 (4.39)	3.28 (4.50)	0.30 (0.05)	0.00 (-1.54 1.54)	-0.91 (-2.40 0.57)	-0.91 (-2.66 0.84)
	Knee frontal RoM [°]	5.03 (2.51)	4.89 (2.04)	4.91 (2.20)	0.65 (0.00)	0.14 (-1.19 1.46)	0.12 (-1.07 1.31)	-0.02 (-1.25 1.22)
	Knee peak adduction moment [Nm/kg]	3.56 (3.59)	3.75 (3.78)	3.87 (3.88)	0.13 (0.10)	-0.19 (-0.53 0.15)	-0.32 (-0.64 0.01)	-0.13 (-0.43 0.18)
	Knee peak abduction moment [Nm/kg]	-1.81 (2.08)	-2.07 (2.35)	-1.95 (2.21)	0.40 (0.04)	0.26 (-0.15 0.66)	0.14 (-0.30 0.59)	-0.11 (-0.56 0.33)
Transverse plane	Knee peak int. rotation [°]	28.46 (11.62)	28.27 (9.45)	29.73 (11.20)	0.79 (0.01)	0.19 (-4.52 4.91)	-1.27 (-6.73 4.19)	-1.47 (-5.79 2.86)
	Knee peak ext. rotation [°]	42.57 (12.65)	43.32 (8.82)	43.39 (11.43)	0.17 (0.00)	-0.76 (-5.87 4.35)	-0.82 (-6.71 5.07)	-0.06 (-3.96 3.83)
	Knee transverse RoM [°]	14.11 (3.97)	15.06 (2.77)	13.66 (3.56)	0.23 (0.07)	-0.95 (-2.75 0.84)	0.45 (-1.50 2.40)	1.40 (0.07 2.73)
	Knee peak internal rotation moment [Nm/kg]	0.95 (0.49)	0.90 (0.46)	0.81 (0.43)	0.30 (0.06)	0.05 (-0.07 0.17)	0.14 (-0.10 0.39)	0.09 (-0.11 0.29)
	Knee peak external rotation moment [Nm/kg]	-1.30 (0.84)	-1.49 (1.09)	-1.49 (1.13)	0.40 (0.07)	0.19 (-0.03 0.42)	0.19 (-0.08 0.46)	0.00 (-0.29 0.29)

Note. P-values as revealed by one-way repeated measures ANOVAs / Friedman tests and effect sizes (η_p^2); significant results ($p < 0.05$) in bold. MD = Mean difference; 95%CI = 95 % confidence intervals; B = baseline; S = short-term; M = medium-term; RoM = Range of Motion.

Table A.7: Results of bracing effects on discrete kinematic and kinetic parameters of the knee joint during the spin turn.

Parameter during spin turn		Base-line	Short-term	Medium-term	ANOVA/ Friedman	B vs. S		B vs. M		S vs. M	
		Mean (SD)	Mean (SD)	Mean (SD)	p (η_p^2)	MD (95% CI)	p	MD (95% CI)	p	MD (95% CI)	p
Sagittal plane	Knee peak extension[°]	11.90 (3.64)	10.02 (2.73)	9.53 (4.49)	0.09 (0.13)	1.88 (0.59 3.17)		2.37 (-0.40 5.15)		0.49 (-1.49 2.48)	
	Knee peak flexion [°]	46.74 (5.15)	45.75 (4.94)	45.75 (6.22)	0.55 (0.02)	0.99 (-0.31 2.28)		0.99 (-1.88 3.85)		0.00 (-2.77 2.77)	
	Knee sagittal RoM [°]	34.84 (4.23)	35.73 (5.35)	36.22 (6.06)	0.35 (0.05)	-0.89 (-2.25 0.47)		-1.39 (-3.89 1.12)		-0.49 (-2.60 1.61)	
	Knee peak extension moment [Nm/kg]	-1.75 (1.80)	-1.89 (1.93)	-1.99 (2.23)	0.87 (0.03)	0.15 (-0.19 0.49)		0.25 (-0.41 0.90)		0.10 (-0.39 0.59)	
	Knee peak flexion moment [Nm/kg]	3.06 (3.35)	3.37 (3.74)	3.19 (3.50)	0.37 (0.08)	-0.30 (-0.71 0.10)		-0.13 (-0.31 0.06)		0.17 (-0.23 0.58)	
Frontal plane	Knee peak adduction [°]	-2.98 (3.65)	-3.30 (5.22)	-2.65 (4.86)	0.75 (0.01)	0.32 (-1.41 2.05)		-0.33 (-2.00 1.34)		-0.65 (-2.55 1.26)	
	Knee peak abduction [°]	2.20 (4.56)	1.56 (4.15)	2.82 (4.66)	0.47 (0.07)	0.64 (-0.89 2.17)		-0.63 (-2.07 0.81)		-1.27 (-2.79 0.26)	
	Knee frontal RoM [°]	5.17 (2.02)	4.86 (1.92)	5.47 (1.97)	0.23 (0.03)	0.32 (-1.04 1.68)		-0.30 (-1.50 0.90)		-0.62 (-1.61 0.37)	
	Knee peak adduction moment [Nm/kg]	0.69 (1.08)	1.44 (1.71)	1.29 (1.87)	0.01 (0.20)	-0.75 (-1.30 -0.20)	0.015	-0.60 (-1.26 0.06)	0.053	0.15 (-0.15 0.45)	0.243
	Knee peak abduction moment [Nm/kg]	-2.63 (1.97)	-2.42 (2.37)	-2.62 (2.52)	0.23 (0.02)	-0.21 (-0.80 0.39)		-0.01 (-0.68 0.66)		0.20 (-0.32 0.71)	
Transverse plane	Knee peak int. rotation [°]	24.49 (11.28)	23.28 (9.98)	25.64 (12.61)	0.59 (0.03)	1.21 (-3.30 5.73)		-1.14 (-6.46 4.17)		-2.36 (-6.68 1.97)	
	Knee peak ext. rotation [°]	46.66 (12.13)	46.02 (9.64)	46.72 (11.57)	0.95 (0.00)	0.65 (-4.70 5.99)		-0.06 (-6.22 6.10)		-0.71 (-4.85 3.43)	
	Knee transverse RoM [°]	22.17 (6.70)	22.74 (4.15)	21.09 (4.60)	0.21 (0.08)	-0.56 (-2.83 1.70)		1.08 (-0.84 3.01)		1.65 (0.06 3.24)	
	Knee peak internal rotation moment [Nm/kg]	1.03 (0.60)	1.45 (0.83)	1.47 (0.93)	0.01 (0.24)	-0.42 (-0.74 -0.10)	0.021	-0.43 (-0.80 -0.07)	0.025	-0.02 (-0.14 0.11)	0.919
	Knee peak external rotation moment [Nm/kg]	-0.71 (0.33)	-0.57 (0.48)	-0.75 (0.69)	0.40 (0.05)	-0.14 (-0.38 0.09)		0.04 (-0.30 0.37)		0.18 (-0.02 0.38)	

Note. P-values as revealed by one-way repeated measures ANOVAs / Friedman tests and Holm-Bonferroni corrected pairwise comparisons and effect sizes (η_p^2); significant results ($p < 0.05$) in bold. MD = Mean difference; 95%CI = 95 % confidence intervals; B = baseline; S = short-term; M = medium-term; RoM = Range of Motion.

Table A.8: Spearman rank correlation coefficients (ρ) with respective p-values for correlations between biomechanical parameters of the step turn and Harris Hip Scores from the hip osteoarthritis group.

Variable	ρ	P
Hip peak extension [°]	-0.26	0.259
Hip sagittal range of motion [°]	0.24	0.294
Hip transverse range of motion [°]	0.36	0.112
Knee peak flexion [°]	-0.02	0.922

Table A.9: Spearman rank correlation coefficients (ρ) with respective p-values for correlations between biomechanical parameters of the spin turn and Harris Hip Scores from the hip osteoarthritis group.

Variable	ρ	P
Hip peak extension [°]	-0.36	0.106
Hip sagittal range of motion [°]	0.19	0.420
Hip peak adduction [°]	0.07	0.752
Hip frontal range of motion [°]	-0.09	0.691
Hip transverse range of motion [°]	0.14	0.531

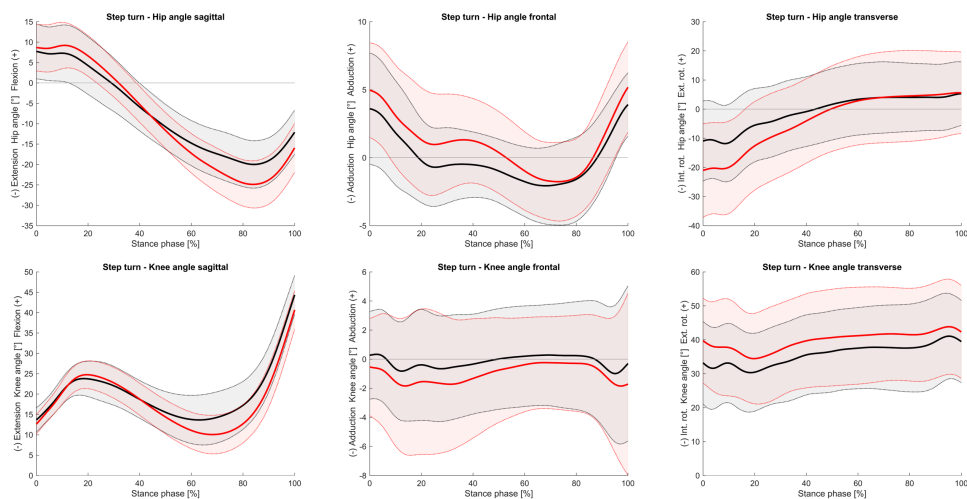


Figure A.28: Hip and knee joint angle time curves for the step turn.

Note. Data presented as mean $\pm$ SD. Black = hip osteoarthritis group at baseline; red = control group.

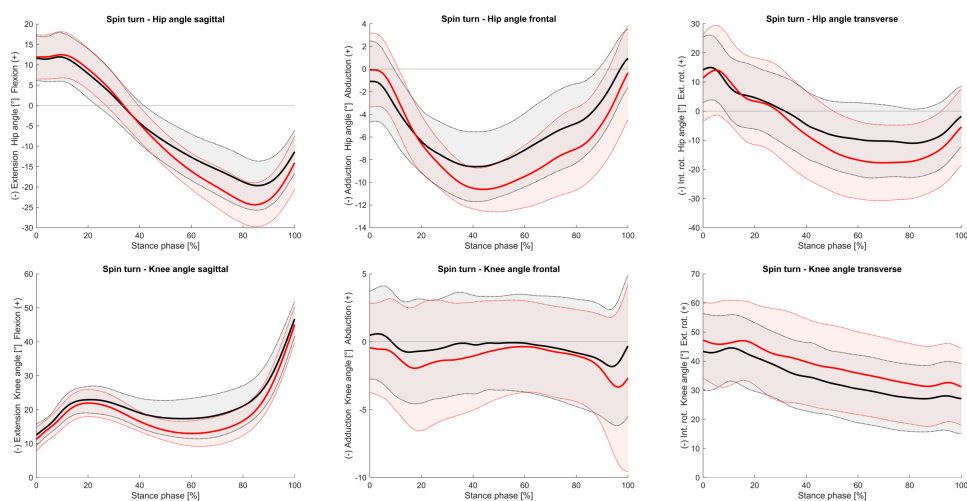


Figure A.29: Hip and knee joint angle time curves for the spin turn.

Note. Data presented as mean $\pm$ SD. Black = hip osteoarthritis group at baseline; red = control group.

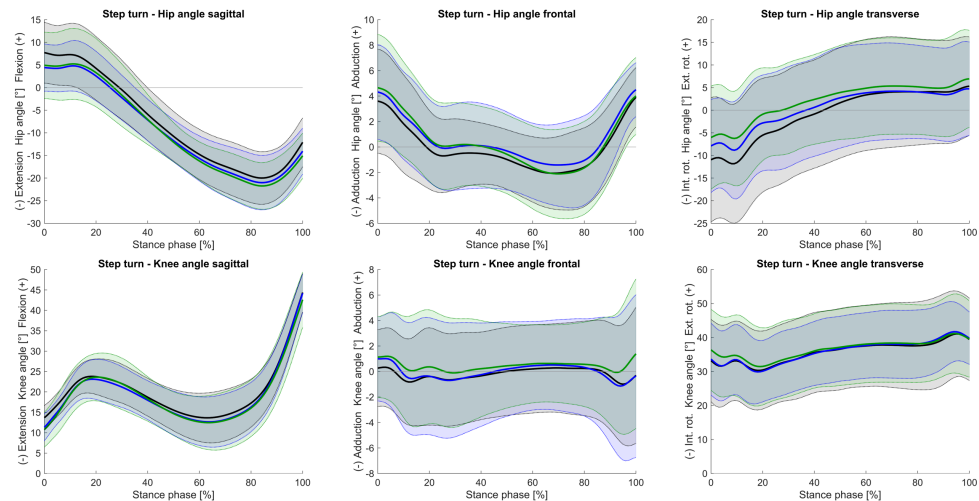


Figure A.30: Hip and knee joint angle time curves for the step turn.
Note. Data presented as mean $\pm$ SD. Black = hip osteoarthritis (HOA) group at baseline; blue = HOA group with short-term bracing; green = HOA group with medium-term bracing.

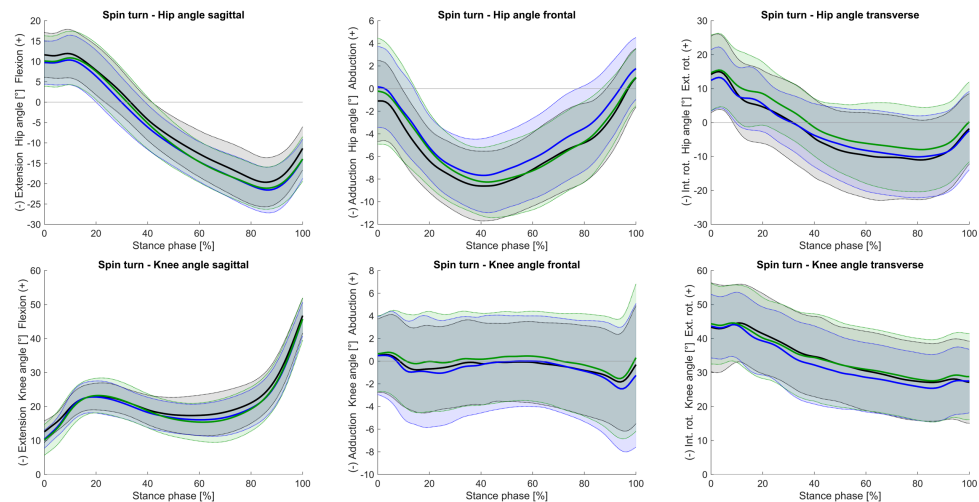


Figure A.31: Hip and knee joint angle time curves for the spin turn.
Note. Data presented as mean $\pm$ SD. Black = hip osteoarthritis (HOA) group at baseline; blue = HOA group with short-term bracing; green = HOA group with medium-term bracing.

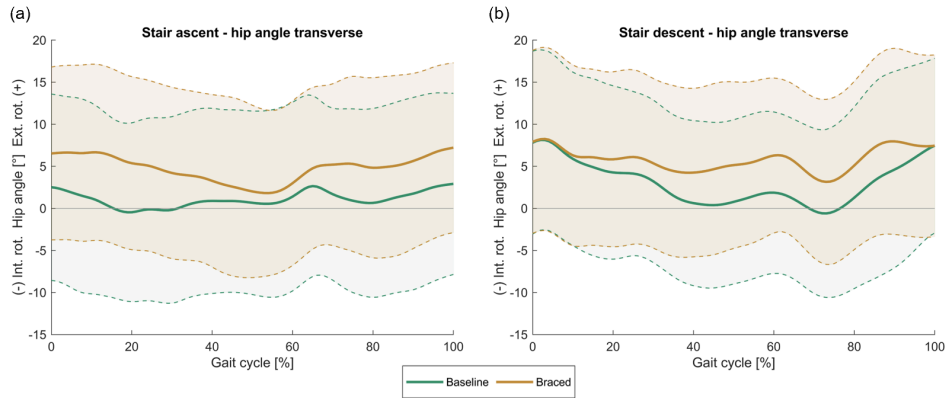


Figure A.32: Transverse plane hip angle [°] across the gait cycle during (a) stair ascent and (b) stair descent.
Note. Data presented as mean $\pm$ SD.

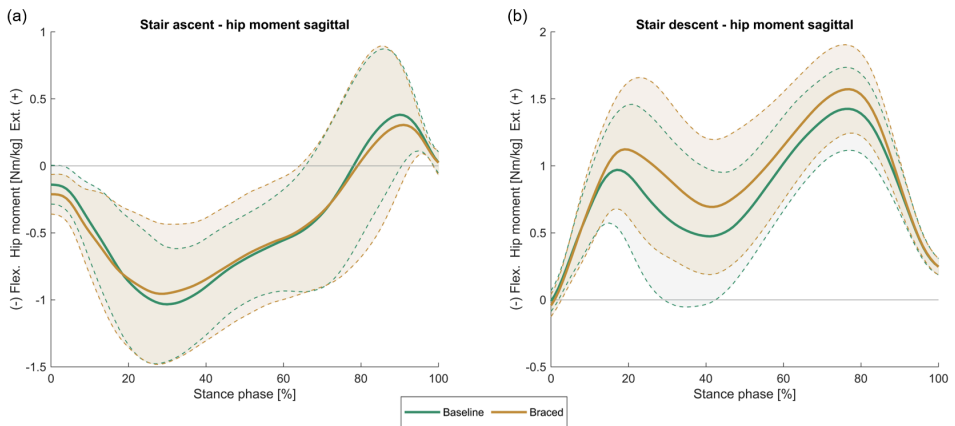


Figure A.33: Sagittal plane external hip moments [Nm/kg] across the stance phase during (a) stair ascent and (b) stair descent.
Note. Data presented as mean $\pm$ SD.

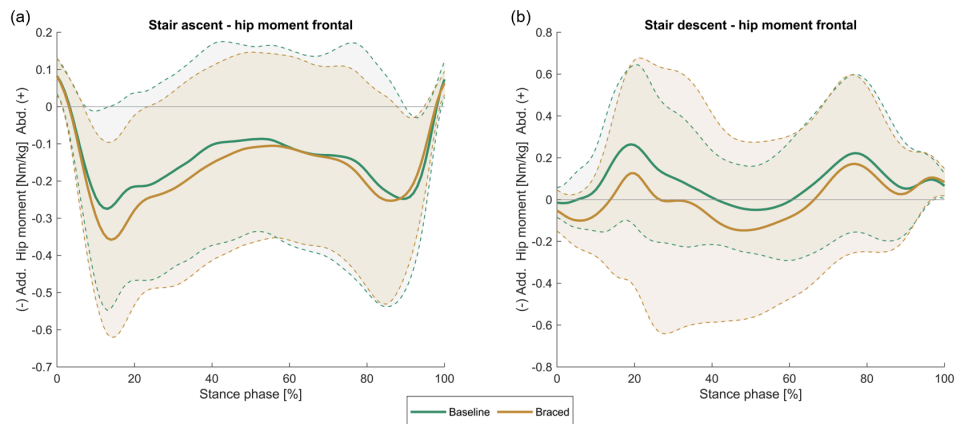


Figure A.34: Frontal plane external hip moments [Nm/kg] across the stance phase during (a) stair ascent and (b) stair descent.
Note. Data presented as mean $\pm$ SD.

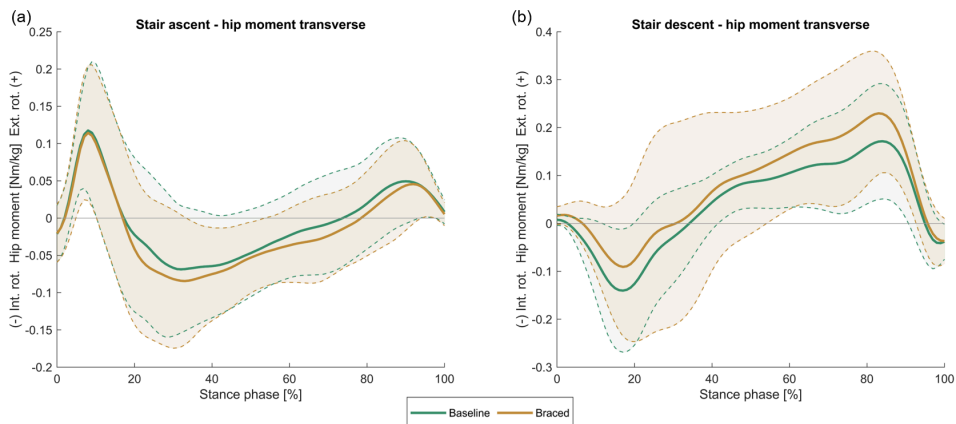


Figure A.35: Transverse plane external hip moments [Nm/kg] across the stance phase during (a) stair ascent and (b) stair descent.
Note. Data presented as mean $\pm$ SD.

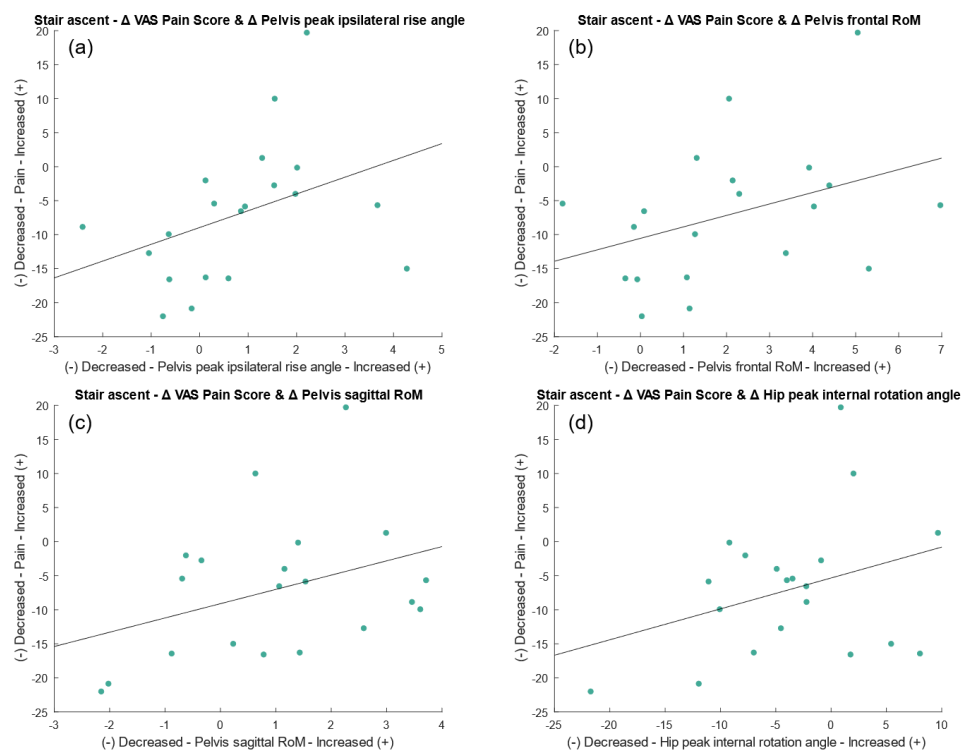


Figure A.36: Scatter plots of Δ VAS pain score versus Δ (a) pelvis peak ipsilateral rise angle, (b) pelvis frontal range of motion (RoM), (c) pelvis sagittal RoM, (d) hip peak internal rotation angle during stair ascent.

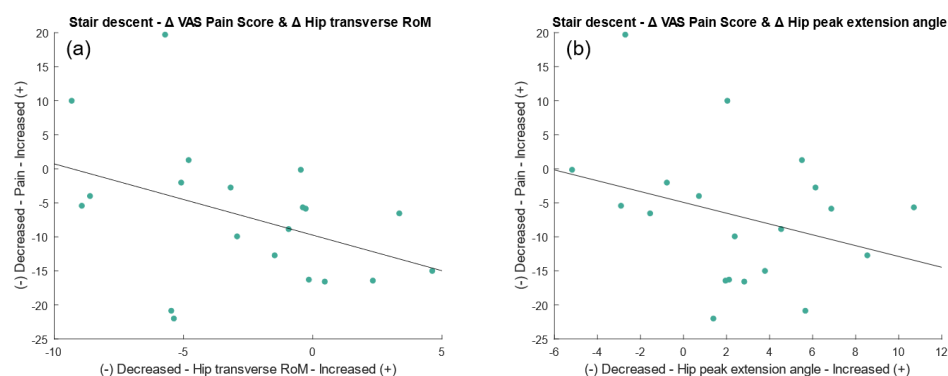


Figure A.37: Scatter plots of Δ VAS pain score versus Δ (a) hip transverse range of motion (RoM) and (b) hip peak extension angle during stair descent.

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