University of Pannonia

Doctoral School of Chemical Engineering and Material Sciences

Submitted for the degree of
Doctor of Philosophy
of the University of Pannonia, Hungary

DOI:10.18136/PE.2026.999

Author: Li Xin
Supervisor(s): Dr. habil. Gusztáv Fekete and Dr. András Kovács

Dissertation Title: Biomechanical Analysis of Walking Stability During Pregnancy and the Fall-Prevention Effects of Negative Heel Shoes

Veszprém
2026

Biomechanical Analysis of Walking Stability During Pregnancy and the Fall-Prevention Effects of Negative Heel Shoes

The thesis was prepared for the award of a doctoral degree (PhD) within the framework of the Chemical Engineering and Material Sciences Doctoral School at University of Pannonia

in the discipline of bio-, environmental-, and chemical engineering sciences

written by: Li Xin

Supervisors: Dr. habil. Fekete Gusztáv and Dr. Kovács András

I recommend the dissertation for acceptance: yes.

……………………….
Dr. habil. Fekete Gusztáv

I recommend the dissertation for acceptance: yes.

……………………….
Dr. Kovács András

I recommend the dissertation for peer review.

……………………….
chair of the DDHC

The PhD-candidate has achieved .……. % at the public debate.

The composition of the Final Examination Committee:
chair:……….……………………………………………………………………………
reviewers:..………………………………………………………………………………
members:……………………………………………………………………………….
Veszprém,

……………………….
chair of the committee

Qualification of degree: …...............................
Veszprém,

……………………….
chair of the UDHC

Contents

Biomechanical Analysis of Walking Stability During Pregnancy and the Fall-Prevention Effects of Negative Heel Shoes

Abstract

During pregnancy, gestational weight gain increases maternal loading, while the added gestational mass is distributed unevenly across body segments, particularly in the anterior-inferior trunk region. These changes may modify the mechanical conditions under which walking is performed and may affect foot structure, gait mechanics, and stability-related biomechanical indicators. However, the mechanical relationships among these factors remain insufficiently clarified. This thesis investigated this problem from three related perspectives: structural response of the foot, walking biomechanics in late pregnancy, and the acute effects of footwear with different heel-toe drops.

The first part of the thesis focused on foot arch deformation during pregnancy. A subject-specific three-dimensional finite element model was used to examine the separate mechanical effects of increased body weight-related loading and reduced passive plantar ligament–fascia stiffness on transverse-arch geometry and internal stress distribution. A full-factorial parametric simulation design was applied, and regression was used as a descriptive sensitivity-analysis tool. Within the examined model, increased loading was more strongly associated with mediolateral widening of the transverse arch and lateral displacement of the arch apex, whereas reduced passive tissue stiffness was more strongly associated with loss of effective arch height and relative curvature. The model also showed increased internal stress in the midfoot, with particularly pronounced increases in the medial cuneiform and cuboid regions. These findings indicate that increased loading and reduced passive tissue support may contribute to pregnancy-related foot deformation through partly distinct mechanical response patterns.

The second part of the thesis examined walking biomechanics in late pregnancy. Walking data were collected from pregnant and non-pregnant women using wearable gait assessment, three-dimensional motion capture, and force-platform measurements. A pregnancy-specific OpenSim musculoskeletal model was then used, with segmental mass parameters and center-of-mass locations adjusted to account for pregnancy-related anthropometric changes. This model adjustment allowed the observed gait adaptations to be interpreted in a pregnancy-specific mechanical context, rather than only through external gait parameters or a generic musculoskeletal model. Induced acceleration analysis was further applied to estimate model-based muscle-force-induced joint angular acceleration patterns during stance. Compared with non-pregnant controls, women in late pregnancy walked with lower speed, shorter stride length, lower cadence, reduced segmental

accelerations, and lower sagittal-plane joint moments. The induced acceleration results showed joint- and muscle-specific differences in the signed model-estimated joint angular accelerations. In the IAA subset, the clearest distal changes occurred at the ankle, where pregnant women showed a smaller gastrocnemius-induced plantarflexion-related acceleration and a smaller tibialis-anterior-induced dorsiflexion-related acceleration. These findings suggest that late-pregnancy walking is characterized by a lower-output and more conservative gait pattern, accompanied by altered muscle-force-induced joint angular acceleration patterns.

The third part examined the acute biomechanical effects of negative-heel shoes, flat shoes, and low-heel shoes in women in late pregnancy. Under self-selected walking-speed conditions, negative-heel shoes led to a slower and shorter-step walking pattern, with longer single-support time than low-heel shoes. They also changed stance-phase loading, with footwear-related differences in vertical and anteroposterior ground reaction forces occurring mainly during specific periods of stance rather than throughout the whole stance phase. In sagittal-plane joint mechanics, negative-heel shoes altered ankle position during late stance and produced only short phase-specific changes in knee angle, whereas ankle and knee joint moments were not significantly different among footwear conditions after correction. Negative-heel shoes were also associated with a smaller peak backward center of mass–center of pressure tilt angle. Taken together, these findings suggest that negative-heel shoes mainly modulated walking organization, lower-limb posture, force timing, and COM–COP alignment, rather than increasing sagittal-plane ankle or knee moment output. This pattern provides acute biomechanical evidence that heel-toe drop can influence stability-related gait indicators in late pregnancy, while further speed-controlled and longitudinal studies are needed to determine whether these changes translate into fall-risk reduction in daily walking.

This thesis shows that pregnancy-related walking biomechanics are not driven by a single factor, but by the interaction between foot structural loading, conservative gait adaptation, and footwear-related mechanical constraints. These findings provide a mechanical basis for pregnancy-specific gait assessment and footwear design, while future studies should verify them using larger models, direct physiological measurements, speed-controlled experiments, and long-term safety outcomes.

Abbreviations

BW: body weight
AP: anteroposterior
C1: medial cuneiform
C3: lateral cuneiform
COP: center of pressure
Cu: cuboid
FE: finite element

IAA: induced acceleration analysis
K: tissue stiffness
SPM: statistical parametric mapping
NHS: negative-heel shoes
SPM1D: one-dimensional statistical parametric mapping
ANOVA: analysis of variance
BMI: body mass index
C2: intermediate cuneiform
COM: center of mass
CT: computed tomography
W: body-weight-related single-foot vertical load
GRF: ground reaction force
CON: control group
ROM: range of motion
ML: mediolateral
SD: standard deviation
RRA: residual reduction algorithm

List of Figures

List of Tables

1. Introduction

1.1 Gestational weight gain and connective tissue mechanical characteristics

1.1.1 Gestational weight gain

Gestational weight gain in pregnant women results from the combined effects of fetal growth, placental development, amniotic fluid accumulation, uterine enlargement, blood-volume expansion, extracellular fluid retention, and maternal tissue accretion. According to the 2010 Institute of Medicine recommendations, the expected total gestational weight gain ranges from 12.5 to 18.0 kg in underweight women, 11.5 to 16.0 kg in women with normal pre-pregnancy body mass index, 7.0 to 11.5 kg in overweight women, and 5.0 to 9.0 kg in women with obesity [1]. These changes are also temporally non-uniform. Gestational weight gain is typically limited in early pregnancy and becomes more sustained from the second trimester onward [1, 2]. The mechanical implication is that gestational weight gain progressively increases maternal loading. At the same time, the added gestational mass is distributed unevenly across body segments. As a result, gestational weight gain and the uneven distribution of added gestational mass may alter segmental loading, the location of the body center of mass, and the mechanical demand placed on the lower limbs and feet. This distinction is important because gestational weight gain differs mechanically from obesity-related weight gain. In pregnancy, by contrast, gestational weight gain consists of multiple components with distinct anatomical locations and mechanical consequences. For a reference gestational weight gain of 12.5 kg, reported contributions at term include approximately 2414 g in the fetus, 540 g in the placenta, 792 g in amniotic fluid, 800 g in uterine tissue, 304 g in mammary tissue, 1267 g in blood, and 1496 g in extracellular fluid [3]. Body-composition studies further indicate that placental mass typically reaches 600–700 g at term, amniotic fluid may accumulate to approximately 3 L, and fetal mass increases rapidly in late gestation after reaching about 1 kg at 28 weeks [4]. Therefore, a considerable proportion of the added mass is fluid or fluid-associated tissue, rather than solid tissue such as fat or skeletal muscle. This composition may influence segmental inertial properties differently from more solid forms of mass gain. Because much of this added mass is located in the anterior-inferior trunk, it may shift the maternal center of mass forward, increase the demand for postural and gait control, and increase the load transmitted through the lower limbs to the feet.

Experimental simulations of pregnancy-related anterior loading in non-pregnant women have shown that adding mass to the anterior trunk can produce measurable changes in lower-limb kinematics and joint moments [5]. However, such simulations do not fully reproduce the progressive, multi-component, and pregnancy-specific uneven distribution of added gestational mass during actual pregnancy, nor do they directly explain how these changes are reflected in plantar loading in pregnant women.

The postural literature similarly suggests that pregnancy is associated with alterations in trunk and pelvic alignment, although the reported direction and magnitude of these changes are not fully consistent across studies. Franklin and Conner-Kerr reported significant increases in lumbar angle and sagittal pelvic tilt from the first to the third trimester [6], whereas Betsch et al. observed increased thoracic kyphosis without a corresponding increase in lumbar lordosis and found no significant change in pelvic position [7]. A systematic review reached a similar conclusion, noting that spinal lordosis and kyphosis often increase during pregnancy, but not uniformly across studies. Taken together, these findings suggest that pregnancy-related postural adaptation is present but methodologically sensitive [8]. Biomechanically, any systematic alteration in trunk-pelvic alignment may change the line of action of body weight relative to the lower-limb joints and thereby affect joint moment demands, postural support strategy, and plantar load transfer during standing and walking.

In addition to structural and mechanical changes, gestational weight gain may also induce neural and neuromuscular control adaptations within the musculoskeletal system. However, these changes should primarily be interpreted as adaptations in sensorimotor integration, postural regulation, and muscle activation strategies, rather than as direct evidence of structural neural alteration. Balance studies provide relatively clear evidence for pregnancy-related changes in sensory dependence and postural control strategy. Butler et al. found that pregnant women showed reduced postural stability during quiet standing and relied more heavily on visual cues, suggesting that when pregnancy-related mechanical demands challenge balance control, pregnant women may depend more strongly on visual information for postural regulation [9]. Recent studies of center-of-pressure behavior and body sway regulation also support this interpretation. Ramachandra et al. found that, under multiple sensory conditions, including restricted visual input, proprioceptive challenge, or altered support-surface conditions, women in late pregnancy showed greater velocity moment and anteroposterior sway velocity than non-pregnant women. This suggests that, when sensory conditions are constrained, the ability to maintain postural stability in late pregnancy may be more strongly challenged [10]. However, the evidence is not entirely consistent. Dumke et al. examined the contribution of sensory integration and segmental body control to postural control during pregnancy using accelerometer-based measures, but did not find significant differences between pregnant and non-pregnant women in postural sway, visual/vestibular/somatosensory control ratios, or segmental balance-control strategies. This finding suggests that evidence regarding pregnancy-related changes in postural control remains partly inconsistent and requires further verification in larger samples and more dynamic tasks [11]. Pregnancy-related neural control changes are also reflected in altered neuromuscular drive and postural muscle activation strategies. As gestational weight gain progresses and anterior loading increases, the control demands required to maintain upright posture and walking stability may no longer be distributed uniformly across all muscle groups, but may instead be redistributed toward the lumbopelvic region and key lower-limb joints. Moreira et al. found that, compared with non-pregnant controls, pregnant women showed lower maximal voluntary isometric activity in the tested postural muscles, but greater lumbar muscle activity during quiet standing, while soleus activity remained relatively stable throughout pregnancy. This result suggests that pregnancy does not simply increase all postural muscle activity in parallel; rather, it may increase the involvement of lumbopelvic muscles to meet the stability demands imposed by greater anterior loading and altered posture [12]. Therefore, pregnancy-related neuromuscular adaptation should be understood as a reorganization of control strategy rather than as a linear increase in the load or activation of a single muscle group. This reorganization of control strategy is also evident during gait. Although overall gait kinematics in pregnancy may remain relatively limited in magnitude, increased kinetic demands at the hip and ankle suggest that the neuromuscular system must adjust joint moments and muscle recruitment to maintain walking mechanics under altered body-mass distribution [13]. Subsequent three-dimensional gait and electromyographic studies further showed that lower-limb joint moment distribution and gluteus maximus activation amplitude differ between pregnant and nulliparous women, indicating that pregnancy alters not only external loading, but also the coordination between muscle activation and joint-level mechanical demand during walking [14]. When walking tasks become more challenging, this adaptation becomes more apparent. For example, during transitions between level and inclined walking surfaces, pregnant women demonstrated slower and wider gait, greater joint flexion, and higher muscle activity, reflecting a cautious gait-control strategy that prioritizes safety and stability [15]. The relevance of proprioceptive control is also supported by intervention evidence. Proprioceptive training has been shown to reduce anteroposterior, mediolateral, and overall postural sway indices in pregnant women, with improvements maintained at follow-up. This supports the interpretation that pregnancy-related balance changes are related, at least in part, to sensorimotor control rather than only to passive structural change or increased body mass [16]. Overall, the available evidence supports interpreting pregnancy-related neural changes as adaptive reorganization of the sensorimotor and neuromuscular control systems. Gestational weight gain, anterior redistribution of body mass, altered pelvic alignment, plantar loading, and joint position changes may collectively modify mechanical demands and sensory input. In response, the nervous system may help maintain stability by redistributing muscle activation, adjusting visual and proprioceptive dependence, modifying postural sway regulation, and adopting a more cautious gait strategy.

1.1.2 Hormonal changes and connective tissue behavior during pregnancy

The mechanical properties of ligaments that stabilize joints have also been reported to be influenced by hormonal regulation during pregnancy. Ligamentous laxity during pregnancy is commonly regarded as an important component of pregnancy-related musculoskeletal change. It is better understood as a complex adaptive process shaped by systemic endocrine remodeling, local tissue characteristics, and region-specific mechanical demands. The hormones most frequently implicated in pregnancy-related connective-tissue adaptation are estrogen, progesterone, and relaxin. These hormones have been associated with changes in collagen turnover, extracellular matrix remodeling, and tissue hydration, and may thereby influence the mechanical behavior of ligamentous and capsular tissues [17-20].

From an endocrine perspective, pregnancy is characterized not by minor hormonal fluctuations but by sustained and substantial systemic change over several months. In a longitudinal study of uncomplicated pregnancies, Schock et al. reported that median estradiol concentrations increased from 2.18 ng/mL in early pregnancy to 20.4 ng/mL in late pregnancy, while progesterone increased from 25.6 ng/mL to 130 ng/mL over the same period [21].

Mechanistically, pregnancy-related ligamentous laxity should not be reduced to the single notion that ligaments simply become compliant. A more defensible interpretation is that pregnancy may increase joint laxity and passive tissue compliance. In mechanical terms, this means that the joint may allow greater movement under load before the ligaments and joint capsule provide restraint. This interpretation is supported by studies reporting increased peripheral joint laxity during pregnancy, although the relationship between this laxity and circulating pregnancy hormones remains inconsistent. Estrogen has been implicated in the regulation of collagen synthesis and degradation, partly through effects on matrix metalloproteinases and related extracellular matrix pathways. Progesterone has been associated with changes in tissue hydration, fluid balance, and viscoelastic properties, which may influence deformation and recovery under load [18, 22]. Relaxin has received particular attention because, through binding to relaxin family peptide receptors, especially RXFP1, it is linked to collagen degradation, collagenase activation, and antifibrotic signaling pathways, and therefore has a plausible mechanistic relationship with increased tissue compliance during pregnancy [19, 20]. Existing reviews further suggest that relaxin-related effects are not limited to the pelvic ligaments but may also involve tendons, ligaments, cartilage, and other collagen-rich musculoskeletal tissues [19]. Accordingly, pregnancy-related ligamentous laxity is better interpreted as a local mechanical expression of broader connective-tissue remodeling rather than as an isolated single-joint phenomenon.

Pregnancy-related changes in passive restraint do not appear to follow a single uniform pattern. Cherni et al. reported that metacarpophalangeal joint laxity increased by approximately 11% from the first to the second trimester and then remained relatively stable until delivery, while the Beighton score was significantly higher in the second trimester [23]. Earlier studies also reported increased peripheral joint laxity or range of motion during pregnancy. Calguneri et al. quantified laxity at the metacarpophalangeal joint of the index finger and assessed generalized joint laxity using a modified Beighton scoring system [24]. Schauberger et al. reported increased laxity in five of seven peripheral joints during pregnancy and postpartum, including knee laxity measured by anterior tibial translation, but found no correlation between this increase and serum relaxin levels [25]. Marnach et al. further reported pregnancy-related changes in wrist laxity and examined their relationship with maternal hormones [26]. These findings indicate that pregnancy-related passive tissue changes may be expressed as greater joint displacement or range of motion under standardized testing conditions. However, increased laxity does not necessarily mean that the material stiffness of every ligament or tendon decreases. Bey et al. found that patellar tendon stiffness did not decrease from early pregnancy to late pregnancy and postpartum, with mean values of 1060 ± 195, 1033 ± 238, and 1064 ± 220 N/mm, respectively. Tendon force and relative strain also remained unchanged. In contrast, patellar tendon rest length increased progressively from 48.2 ± 3.3 mm in early pregnancy to 49.3 ± 3.8 mm in late pregnancy and 50.6 ± 3.4 mm postpartum [27]. This suggests that pregnancy-related laxity may occur through changes in rest length, slack length, or viscoelastic behavior, rather than through a universal reduction in tissue stiffness. Animal evidence summarized by Bey et al. provides a similar caution [27]. Hart et al. reported no pregnancy-related reduction in the structural, material, or viscoelastic properties of the rabbit medial collateral ligament, while Rundgren reported no general reduction in rat posterior cruciate ligament or tendon stiffness during gestation [28, 29]. Therefore, pregnancy-related connective-tissue changes should not be interpreted simply as a uniform decrease in ligament or tendon stiffness. The foot is not a single-ligament structure, but a mechanically integrated load-bearing system composed of the longitudinal and transverse arches, the plantar fascia, short plantar ligaments, tarsometatarsal joints, metatarsophalangeal joints, and surrounding soft tissues. Therefore, increased loading and possible changes in passive tissue restraint should be considered as two distinct but concurrent mechanisms. Gestational weight gain increases the external load applied to the foot, whereas pregnancy-related connective-tissue changes may alter the passive restraint provided by ligaments, the plantar fascia, and capsular tissues. The final mechanical response of the foot therefore depends on the interaction between external loading, tissue properties, and foot geometry. Its passive stability depends on the interaction among geometry, material properties, and dynamic loading. Previous studies have shown that measurable foot changes do occur during pregnancy, including increases in metatarsophalangeal girth of approximately 9–28 mm, as well as increases in foot length, foot width, and plantar contact area, reductions in arch-height-related indices, and increases in rearfoot eversion; some of these changes may persist postpartum [30, 31]. These observations should not be interpreted as evidence that all foot ligaments become uniformly more compliant during pregnancy. As foot geometry changes, the relative positions of bones and ligament attachment sites may also change. Some passive structures may be elongated and placed under greater tensile demand, whereas others may become relatively shortened or unloaded, depending on their anatomical location and line of action. For example, flattening of the foot arch may stretch the plantar fascia and increase its tensile demand, while pregnancy-related changes in tissue hydration, collagen turnover, and viscoelastic behavior may simultaneously reduce the effective material stiffness or passive restraint of some connective tissues. Therefore, pregnancy-related foot deformation is better interpreted as the result of an interaction between gestational weight gain-related loading, altered foot geometry, and possible changes in connective-tissue material behavior, rather than as a uniform change in ligament stiffness. However, these observations primarily reflect system-level structural and loading adaptations and should not be interpreted as direct evidence that foot-specific ligament stiffness has been quantitatively shown to decline during pregnancy. The current evidence supports a cautious interpretation. The marked endocrine remodeling of pregnancy, together with in vivo evidence of peripheral joint laxity, makes adaptive changes in distal passive support tissues mechanistically plausible. However, the extent to which such changes are expressed in the foot as quantifiable reductions in ligament stiffness, plantar fascial stiffness, or passive structural stability remains insufficiently established. Direct, systematic, and tissue-specific evidence for foot connective-tissue mechanical change during pregnancy is still limited.

Taken together, current evidence supports the plausibility of pregnancy-related changes in passive tissue behavior, but the extent to which such changes are expressed in the foot remains unclear. This uncertainty is important because the foot is a load-bearing structure in which geometry, tissue support, and external loading interact directly. Clarifying this issue is therefore necessary for a more specific mechanical interpretation of pregnancy-related foot deformation.

1.2 Foot structural adaptation and walking-related biomechanical changes

1.2.1 Pregnancy-related changes in foot structure

The foot forms the distal load-bearing interface between the body and the ground. It is composed of 26 bones, multiple joints, ligaments, intrinsic and extrinsic muscles, the plantar fascia, and specialized plantar soft tissues, which together support load transfer, shock attenuation, and forward progression during standing and walking. The foot is the distal structure through which ground reaction forces enter the body, the effects of gestational weight gain, the anterior location of added gestational mass, and possible changes in passive support properties may first be reflected at this distal load-bearing interface. The additional mass due to pregnancy not only increases the total mechanical load on the foot, but may also alter the way in which that load is transmitted within the foot. Previous studies have shown that measurable changes in foot dimensions, arch-related parameters, and plantar load distribution can occur during pregnancy [32, 33]. Therefore, the foot can be regarded as an important distal interface for pregnancy-related load transfer and structural adaptation. Pregnancy may increase not only the overall load but also alter the internal pathway by which load is transferred across the rearfoot, midfoot, and forefoot, thereby helping to explain subsequent foot deformation, redistribution of plantar pressure, and walking-related biomechanical adjustments.

Measurable changes in foot structure have been reported during pregnancy, although the magnitude, timing, and persistence of these changes are not fully consistent across studies, likely reflecting differences in study design, measurement methods, gestational timing, sample characteristics, and selected outcome variables. Longitudinal studies and recent reviews generally indicate that increases in foot length, width, and volume are among the more commonly reported findings, and that some of these changes may persist into the postpartum period. One longitudinal study of 30 pregnant women reported that, from 20 to 38 weeks of gestation, mean foot length increased by 0.86 cm (3.6%), foot width by 0.25 cm (2.6%), and rearfoot surface area by 18.36 cm^2 (11.9%), while arch height decreased by 0.52 cm (24.2%). The authors further reported that body mass accounted for more than 90% of the variation in these foot dimensions, suggesting that progressive mechanical loading may be an important contributor to pregnancy-related foot morphological change [34].

Other prospective studies have also documented pregnancy-related changes in foot structure and posture, including increased foot length, reduced arch height, and a tendency toward greater pronation in late pregnancy, although the magnitude and reversibility of these changes vary across studies. Increased mechanical loading has been proposed as one of the principal contributors to these changes. As gestational weight gain progresses, the foot is exposed to greater sustained loading, which is likely to contribute to progressive changes in foot dimensions and arch-related morphology. Fluid retention and soft-tissue swelling may also play a role. Review evidence indicates that increases in foot volume are among the more consistent observations during pregnancy, suggesting that changes in soft-tissue condition may contribute to altered foot shape [32]. Accordingly, increases in foot dimensions during pregnancy should not be interpreted as reflecting osseous or ligamentous change alone.

However, not all studies have reported clear and sustained increases in every dimensional outcome. For example, a longitudinal study assessing women at 9–13 weeks of gestation, 32–35 weeks of gestation, and 4–6 weeks postpartum found no significant overall change in footprint length, although a slight increase was observed in late pregnancy and tended to recover after delivery; the same study also reported a shift toward a more pronated foot posture during pregnancy, with a return toward a more neutral posture postpartum [35]. These differences suggest that inconsistencies across studies may reflect variation in measurement outcomes, assessment timing, sample characteristics, and methodology. Current evidence indicates that pregnancy can alter foot length, width, volume, arch-related measures, and aspects of overall foot morphology. These changes are most plausibly related to a combination of increased mechanical loading, fluid retention, and changes in soft-tissue condition, but the extent and duration of change are not uniform across studies. Pregnancy-related foot adaptation should therefore be understood as a broadly consistent yet methodologically heterogeneous process.

1.2.2 Changes in arch geometry during pregnancy

The foot arch may be understood as a load-sharing system in which the plantar fascia, plantar ligaments, and intrinsic and extrinsic musculotendinous structures interact to modulate foot stiffness during weight-bearing activities. Experimental evidence has shown that restricting longitudinal arch compression during running increases metabolic cost, supporting the view that normal arch deformation contributes to locomotor economy [36-38]. Biomechanical studies further indicate that the arch contributes to elastic energy storage and return during gait. Although these findings are not specific to pregnant populations, they suggest that even relatively small changes in arch structure may have meaningful mechanical consequences.

In pregnancy-related research, reductions in arch-related measures have been reported repeatedly, particularly in the medial longitudinal arch. One longitudinal study of 30 pregnant women found that, from 20 to 38 weeks of gestation, arch height decreased by 0.52 cm (24.2%), while foot length increased by 0.86 cm (3.6%) and foot width by 0.25 cm (2.6%) [34]. The same study reported that body mass accounted for more than 90% of the variation in foot dimensions during pregnancy, suggesting that arch change may occur in parallel with increased mechanical loading and broader changes in foot morphology rather than as an isolated geometric event. Other longitudinal studies have reported changes in the same direction, although the absolute magnitude of change has often been smaller or the measurement methods have differed. In a longitudinal study of 23 women assessed at 9–13 weeks of gestation, 32–35 weeks of gestation, and 4–6 weeks of postpartum, foot posture became more pronated during pregnancy and tended to return toward neutrality after delivery, while footprint length showed only a slight late-pregnancy increase and no significant overall change [35]. These findings suggest that pregnancy-related arch and posture changes can be detected, but that their expression depends in part on the outcome measure used. Prospective postpartum data further suggest that at least some arch changes may persist beyond delivery. In a cohort of 49 women assessed in early pregnancy and again 19 weeks postpartum, standing arch height index decreased, arch drop increased, and arch rigidity index decreased significantly, supporting the view that pregnancy may be associated with a lasting reduction in arch height and rigidity in at least some women. The authors also noted that these changes were most evident after a first pregnancy [31].

Taken together, current evidence indicates that arch-related changes during pregnancy are reported across different study designs, but their magnitude depends substantially on the variables and methods used. The medial longitudinal arch remains the most consistently described site of change, yet these changes often occur together with increased foot width, greater pronation, and broader alterations in foot morphology.

1.2.3 Changes in plantar loading and pressure redistribution during pregnancy

Changes in foot structure during pregnancy are reflected not only in alterations in foot dimensions and arch-related measures, but also in changes in plantar pressure distribution. Evidence from both longitudinal and comparative studies suggests that these plantar pressure changes are not best described as a uniform increase across all plantar regions. Rather, they are more commonly characterized by a redistribution of load among the rearfoot, midfoot, and forefoot [30, 39, 40]. In late pregnancy, gestational weight gain, anterior mass gain, postural adjustment, edema, and changes in arch structure and passive support properties likely together alter plantar contact geometry and regional pressure patterns [8, 31]. Recent reviews have similarly noted that pregnancy is often associated with increased forefoot pressure, prolonged forefoot loading time, and altered plantar load distribution, although the specific regional patterns reported across studies are not entirely consistent [8, 41].

One of the clearest quantitative findings concerns the midfoot. A study comparing women at 38 weeks of gestation with non-pregnant controls found significantly higher mean midfoot pressure in the pregnant group (115.5 kPa vs 95.4 kPa, $p = 0.001$) [42]. In the subgroup reassessed 4 months postpartum, both mean and peak midfoot pressures declined markedly: mean midfoot pressure decreased from 111.9 kPa in late pregnancy to 66.2 kPa postpartum, corresponding to a reduction of approximately 40.8% ($p < 0.001$), and peak midfoot pressure decreased from 184.0 kPa to 108.3 kPa, corresponding to a reduction of approximately 41.1% ($p < 0.001$) [42]. These data indicate that at least part of the increase in midfoot loading observed in late pregnancy is reversible after delivery and is therefore more consistent with pregnancy-specific structural and loading changes than with fixed between-subject differences. This interpretation is also consistent with reports that some pregnancy-related changes in foot structure persist beyond delivery, with arch height and arch stiffness remaining below pre-pregnancy levels at 19 weeks postpartum in longitudinal follow-up studies [31].

Observations from different populations support the same general interpretation. Existing studies indicate that plantar loading during pregnancy changes in a region-specific rather than a uniform manner. In a prospective study of 70 primigravid women followed from the 12th week of gestation to 6 weeks postpartum, static contact area increased from 54.5 cm^2 at 12 weeks to 62.0 cm^2 at 32 weeks and then declined to 56.5 cm^2 at 6 weeks postpartum ($p < 0.001$), while average plantar pressure increased from 983 g/cm^2 to 1112 g/cm^2 across the same period ($p < 0.001$). In the same cohort, static hindfoot pressure in the dominant foot increased from 1845 g/cm^2 at 12 weeks to 2219 g/cm^2 at 32 weeks and remained elevated at 1977 g/cm^2 at 6 weeks postpartum ($p = 0.004$), whereas forefoot pressure also increased significantly over time, for example from 643 g/cm^2 to 887 g/cm^2 on the left side ($p < 0.001$) [40]. These findings suggest that total plantar contact area changes only modestly, whereas pressure increases are not synchronized across regions. Such region-specific pressure redistribution may reflect the combined influence of longitudinal arch lowering and transverse arch widening, although most available plantar-pressure studies do not directly distinguish the relative contribution of these two arch components. Dynamic studies point in the same direction. In late pregnancy, Bertuit et al. found no significant differences across months 6 to 9 of gestation, but clear differences between pregnant and control groups: compared with controls, pregnant women showed a 41% increase in lateral midfoot contact area, a 2% increase in lateral midfoot peak pressure, a 1% increase in medial midfoot peak pressure, and reduced contact area at the rearfoot and forefoot, including decreases of 10% at the medial rearfoot and 8% and 13% at the medial and lateral forefoot, respectively. The time to peak plantar pressure was also delayed by a mean of 0.10 ± 0.03 s over the foot, with the largest increases at the midfoot (77% laterally and 43% medially) [39] . Taken together, these results are more consistent with a redistribution of load from the rearfoot and forefoot toward the midfoot, rather than a generalized increase in pressure across the entire foot. There is also some evidence that these loading changes may be clinically relevant. In a study of 35 third-trimester pregnant women with foot pain and 35 age- and BMI-matched overweight controls, pregnant women showed higher static forefoot pressure on the dominant side (45% vs 38%, $p < 0.05$), higher dynamic dominant forefoot peak pressure (34.0 ± 8.5 vs 32.6 ± 14 kPa, reported as significantly different), and longer forefoot contact times on both sides (0.84 ± 0.3 s and 0.87 ± 0.2 s vs 0.72 ± 0.1 s in controls; both $p < 0.05$) [43]. Importantly, dynamic pain scores were weakly correlated with forefoot contact time on the left ($r = 0.346$, $p = 0.014$) and dominant ($r = 0.314$, $p = 0.027$) sides, suggesting that prolonged forefoot loading may contribute to foot discomfort in late pregnancy.

Taken together, the available evidence suggests that plantar loading during pregnancy is characterized more by a redistribution of load across plantar regions than by a uniform increase in pressure over the whole foot, and this pattern is generally more evident during walking than during quiet standing. Although studies do not report entirely consistent findings for the midfoot, forefoot, contact time, or local pressure variables, these differences are likely related to variation in testing tasks, plantar masking approaches, and the outcome variables selected. Overall, plantar pressure findings indicate that pregnancy-related foot adaptation involves not only morphological change but also a change in the way load is distributed across the foot. At the same time, pressure data alone cannot determine whether these changes are driven primarily by increased external loading or by structural changes in arch configuration and passive support properties. For this reason, pregnancy-related foot deformation is best interpreted together with evidence on foot structure and the underlying mechanical mechanisms.

1.2.4 Mechanical interpretation of foot deformation

Although pregnancy-related changes in the foot have been described with reasonable consistency, their underlying mechanical drivers remain insufficiently distinguished. Broadly, two main explanations have been proposed in the existing literature. The first is an increase in external loading associated with gestational weight gain and the anterior distribution of the added gestational load. Within this framework, foot deformation is interpreted primarily as a load-driven response, whereby gestational weight gain and a forward shift in the center of mass (COM) impose greater mechanical demands on the plantar soft tissues, ligaments, and arch structures. The second explanation concerns change in passive support capacity. From this perspective, if the plantar fascia, ligaments, and related tissues become more compliant during pregnancy, the foot may deform more readily even under similar external loading conditions. The difficulty is that most existing studies describe the co-occurrence of these factors during pregnancy rather than distinguishing their individual contributions. When reductions in longitudinal and transverse arch height, increases in foot width, and alterations in plantar loading are observed, it often remains unclear how much of these responses should be attributed to increased load, how much to reduced tissue restraint, and how much to their interaction. This distinction is important because different mechanisms imply different interpretations of the same observed changes. The distinction is also relevant to intervention. If pregnancy-related foot deformation is driven predominantly by increased external loading, strategies that modify loading conditions may be particularly important, such as postural adjustment, load redistribution, or footwear designs that improve load transfer during standing and walking. By contrast, if reduced passive support plays a substantial role, interventions that enhance structural support or improve load sharing may be more appropriate, such as orthotic devices or shoes providing more effective plantar support. In this sense, distinguishing load-related effects from support-related effects is not only a theoretical issue, but also relevant to the biomechanical rationale underlying different supportive strategies. Pregnancy-related foot deformation is therefore better interpreted as a set of structural responses that may arise from partly distinct mechanical sources rather than as a single undifferentiated phenomenon. Lowering of the longitudinal arch, transverse widening, and redistribution of plantar load do not necessarily result from the same driving factors. Increased external loading is more likely to increase overall compression and pressure, whereas reduced passive support may allow greater deformation under comparable loading conditions. If so, the final pattern of deformation would depend on the interaction between load magnitude and tissue restraint rather than on either factor alone. Accordingly, the key unresolved issue is not whether pregnancy alters foot structure, but which mechanical variables primarily drive the observed deformation patterns. This unresolved mechanistic question provides a direct rationale for the finite element component of the present thesis. By examining, within a common modeling framework, the effects of gestational weight gain-related loading and reduced passive support separately–the study aims to determine whether these two factors produce distinguishable deformation patterns and thereby to provide a more specific mechanical interpretation of pregnancy-related foot adaptation.

1.3 Walking adaptations during pregnancy: observed findings and unresolved mechanisms

1.3.1 Spatiotemporal adaptations during pregnancy walking

Walking is one of the most common forms of physical activity during pregnancy. Among the various gait adaptations reported in the literature, changes in spatiotemporal parameters have been studied most frequently because they provide a relatively direct description of how walking is reorganized as gestation progresses. The added gestational mass is not evenly distributed across body segments, and much of it is located in the anterior trunk region. This may shift the COM and require adjustments in gait pattern. Previous studies have focused mainly on basic characteristics of walking such as walking speed, step or stride length, step width, gait cycle time, cadence, and single- and double-support time. Early reviews of longitudinal evidence suggested that wider steps, shorter strides, and longer support times are among the more commonly reported features of gait adaptation during pregnancy, although the magnitude and statistical significance of these changes have not been consistent across studies [44].

Longitudinal studies indicate that gait adaptation during pregnancy does not occur uniformly across all spatiotemporal variables. Branco et al. reported that, in 22 pregnant women assessed in mid- and late pregnancy and compared with non-pregnant controls, stride length decreased from 1.260 ± 0.098 m to 1.234 ± 0.088 m, left and dominant step length decreased from 0.630 ± 0.051 m and 0.630 ± 0.049 m to 0.616 ± 0.044 m and 0.618 ± 0.045 m, respectively, and double-support time increased, whereas walking speed and step width did not change significantly between these two stages [45]. Another longitudinal study following 30 women across the first, second, and third trimesters found no significant longitudinal effect in the conventional spatiotemporal variables, but reported an increased base of support and a positive association between late-pregnancy physical activity level and both walking speed and stride length [46]. Together, these findings suggest that gait adaptation during pregnancy is real, but that not all variables change in the same way or at the same time.

At the same time, the literature remains heterogeneous. Although many studies support the presence of pregnancy-related gait adjustment, the reported magnitude of change in walking speed, stride length, step width, gait cycle time, and double-support time is not fully consistent, and even the statistical significance of some variables varies across studies. This inconsistency likely reflects differences in gestational stage, sample characteristics, task conditions, and methodological approaches. Previous reviews have also noted that the number of longitudinal gait studies in pregnant populations remains limited, making it difficult for any single study to capture the overall pattern of adaptation [44, 47].

For this reason, our earlier systematic review and meta-analysis was undertaken to synthesise the available reports. A total of 21 studies were included and the pooled findings showed a consistent overall direction of gait adaptation during pregnancy at the population level. Walking speed decreased substantially (Hedges' g = −0.55), stride length decreased (Hedges' g = −0.29), and stride width increased (Hedges' g = 0.45). Gait cycle time and double-support time also increased, with effect sizes of 0.38 and 0.41, respectively. Meta-regression further indicated that advancing gestational age was associated with greater stride-length reduction and stride-width increase. However, between-study heterogeneity remained substantial for walking speed, stride length, and double-support time, whereas stride-width findings showed comparatively good consistency across studies.

Taken together, changes in spatiotemporal gait parameters during pregnancy show a recognisable overall pattern at the population level, although not all variables change to the same extent across studies. Among these parameters, stride shortening and increased stride width appear to be the most consistent findings, whereas walking speed, gait cycle time, and double-support time, although directionally similar overall, appear more sensitive to individual and methodological variation. Interpretation of pregnancy-related gait adaptation therefore requires attention not only to the general population-level trend, but also to the heterogeneity observed across individual studies.

1.3.2 Kinematic and kinetic changes during walking

The spatiotemporal gait adaptations described above are accompanied by changes in joint motion and mechanical loading during pregnancy walking. Compared with external spatiotemporal measures such as walking speed, stride length, and step width, kinematic analysis provides a more direct description of how posture and joint motion are reorganized during gait, whereas kinetic analysis helps to determine whether these adjustments are accompanied by changes in joint moments, mechanical work, and load distribution.

Pregnancy-related gait adaptation also involves important proximal changes in the pelvis–hip system. Longitudinal studies provide direct evidence for such changes. In a repeated-measures study of 20 women assessed in early, middle, and late pregnancy, peak anterior pelvic tilt during walking increased from 8.5 ± 4.9° in early pregnancy to 10.1 ± 3.5° in mid-pregnancy and further to 13.8 ± 5.2° in late pregnancy ($p = 0.005$), whereas the range of pelvic lateral inclination decreased from 7.5 ± 2.0° to 6.3 ± 1.8° ($p = 0.011$). In the same study, peak pelvic inclination was negatively correlated with peak trunk flexion, suggesting that pelvic changes occurred together with adjustments in upper-body posture. Because these pelvic variables were derived from skin-mounted motion-capture markers, they should be interpreted cautiously, as marker placement and soft-tissue artefact may influence the accuracy of pelvic tilt estimation during pregnancy[48]. Similarly, Branco et al. reported that from the second to the third trimester, dominant hip sagittal-plane motion changed significantly, with reduced first-peak hip extension and an increase of more than 5° in second-peak hip flexion. Taken together, these findings suggest that proximal adaptation during pregnancy is not expressed simply as a uniform increase or decrease in joint range of motion, but rather as a redistribution of posture and joint motion within the gait cycle [45].

Other longitudinal studies indicate that pelvic motion may change with gestational progression, although the pattern is not entirely consistent across planes or stages of pregnancy. Gilleard's longitudinal study showed progressive reductions in transverse- and frontal-plane pelvic motion as pregnancy advanced, together with increased step width and reduced stride length [49]. More recent work has further suggested that the most evident changes in the lumbopelvic–hip system may be seen in selected peak variables rather than in a generalized increase or decrease in total excursion. This may help explain why some studies report limited overall changes in pelvic or hip range of motion despite clear alterations in posture and gait organization.

Inter-individual variability also appears to be substantial. In the classic study by Foti et al., overall lower-limb kinematic changes during pregnancy were described as relatively limited, yet peak anterior pelvic tilt still increased by approximately 4° on average [13]. At the same time, the individual responses varied widely, indicating that proximal adaptation is not uniform across subjects [13]. Pain status may further influence the observed results. Bagwell et al. reported that pregnant women with higher low back or pelvic girdle pain scores showed altered gait biomechanics and muscle activation compared with those with lower pain scores, indicating that some of the proximal restriction reported in pregnancy may be more closely related to pain-associated functional limitation than to pregnancy alone [50].

Compared with the kinematic evidence, the kinetic literature provides more consistent support for altered proximal mechanical demand during pregnancy. Foti et al. noted that, despite relatively small overall kinematic changes, several kinetic variables increased significantly during pregnancy, indicating greater mechanical demand on the hip abductors and extensors during walking [13]. This suggests that proximal adaptation does not necessarily appear first as a large change in movement amplitude, but may instead be expressed earlier as an increase in joint moment requirements and muscular loading. Later work has similarly suggested that pregnancy is associated with redistribution of lower-limb mechanical roles, with changes in hip, knee, and ankle moments and in the relative contribution of individual joints to mechanical work across stance. These findings support the view that the significance of hip-related adaptation during pregnancy lies not only in whether joint excursion changes, but also in how the pelvis–hip system contributes to pelvic control, single-limb support, and overall mechanical task distribution during walking. Systematic review evidence likewise indicates that these proximal changes should be interpreted within a broader framework of altered stability demands, posture, and external loading during pregnancy [8]. Pregnancy-related changes in the pelvis–hip system are better understood not as a single passive consequence of body enlargement, but as part of a broader reorganization of gait mechanics shaped by increased stability demands, redistribution of load, and, in some individuals, pain-related functional constraints.

The knee joint also shows adjustment during walking in pregnancy. The most commonly reported findings are increased peak knee flexion and reduced peak extension [8]. Longitudinal evidence suggests that these changes are more likely to occur at specific phases of the gait cycle than as a generalized alteration in overall knee flexion. In a repeated-measures study comparing women in the second and third trimesters of pregnancy with non-pregnant controls, Branco et al. reported a significant increase of approximately 1.2°–1.3° in peak left-knee flexion during the swing phase between the second and third trimesters of pregnancy[45]. These findings suggest that knee adaptation during pregnancy may be expressed as selective phase-specific reorganization within the gait cycle rather than as a uniform change throughout the entire gait cycle.-Kinetic evidence further suggests that pregnancy-related gait adaptation is not limited to visible kinematic change. Foti et al. reported that, although overall gait kinematics remained relatively unchanged during pregnancy, several lower-limb kinetic variables increased significantly, including a ~38% increase in hip abduction moment (73 vs. 53 Nm), a ~19% increase in hip extension moment (44 vs. 37 Nm), and a ~9% increase in ankle plantarflexion moment (101 vs. 93 Nm) [13]. These findings indicate increased mechanical demand on the hip abductors, hip extensors, and ankle plantarflexors during walking.

In walking during pregnancy, the ankle joint and the ankle-foot complex represent a key site of distal adaptation. Review evidence has generally suggested that the most commonly reported kinematic findings at the ankle during pregnancy include reduced plantarflexion, altered inversion-eversion characteristics, and changes in foot progression angle [8]. Longitudinal studies further indicate that these changes are expressed mainly in the sagittal and frontal planes. In a three-dimensional gait analysis of 22 pregnant women assessed in mid and late pregnancy and compared with 12 non-pregnant women, Branco et al. found that dominant ankle plantarflexion decreased significantly from the second to the third trimester, representing one of the clearest distal kinematic changes observed in that study [45]. This finding suggests that ankle function in late pregnancy is modified during terminal stance or late support, but does not indicate that all ankle variables change in parallel. Mei et al. followed 16 women longitudinally in the second trimester, third trimester, and at 4 months postpartum, and similarly identified clear distal changes: peak ankle eversion changed from $-12.07 \pm 2.89°$ in the second trimester to $-7.07 \pm 2.18°$ in the third trimester, before recovering to $-17.76 \pm 4.02°$ postpartum; in parallel, mean plantar pressure distribution and the trajectory of the center of pressure also changed [51]. These findings indicate that changes in ankle-foot motion in late pregnancy occur together with modifications in the plantar loading pathway, suggesting that distal kinematic adaptation is not an isolated angular fluctuation but is more likely linked to reorganization of loading conditions.

This interpretation is also supported by changes in foot placement. Previous reviews have noted an increase in toe-out angle during pregnancy, indicating a more externally rotated foot placement during walking [8]. Although this finding has not been reported uniformly across all studies, its significance is not limited to foot orientation itself. A more externally rotated foot placement changes the relationship between the foot and the ground and may therefore alter frontal- and transverse-plane loading and motion at the ankle. Accordingly, although ankle kinematic findings appear less consistent overall than proximal variables such as anterior pelvic tilt, this does not imply that distal adaptation is of lesser importance. Rather, it suggests that changes in the ankle-foot complex are more sensitive to walking speed, gestational stage, sample characteristics, and measurement methods, and that their expression is more task-dependent. Compared with the kinematic findings, the kinetic evidence for altered ankle function is more consistent. Foti et al. reported that, although overall gait kinematics during pregnancy changed little, kinetic variables at the hip and ankle increased significantly ($p < 0.05$) [13]. This suggests that distal adaptation during pregnancy does not necessarily first appear as a large change in joint excursion, but may instead emerge earlier as an increase in mechanical demand. In other words, even when observable changes in ankle motion are limited, its functional role in support and propulsion may already have been altered. This point was further clarified by Bagwell et al. Compared with nulliparous controls, women in the second trimester already exhibited greater ankle negative work ($p = 0.023$), together with a larger relative contribution of the ankle and a smaller relative contribution of the hip to negative work distribution. By the third trimester, pregnant women showed greater ankle plantarflexion moment ($p < 0.001$), greater ankle dorsiflexion moment ($p = 0.021$), greater ankle negative work ($p = 0.006$), and greater ankle total work ($p = 0.002$), while overall lower-limb positive work and total work also increased [14]. The same study further showed that, in the distribution of negative work during stance, the ankle accounted for approximately 46.6% in the second trimester, compared with 37.3% in controls, whereas the corresponding hip contribution was approximately 28.9% in pregnant women and 36.1% in controls. This indicates that the ankle had already assumed a greater proportion of energy absorption by mid-pregnancy. By the third trimester, although the relative distribution of negative work no longer differed significantly from controls, ankle mechanical demand and overall joint moments of the lower limb increased further [14]. Taken together, these findings support the view that pregnancy gait involves a partial shift in mechanical task demand from proximal to distal segments, with the ankle emerging as one of the principal distal contributors to this redistribution.

Existing evidence indicates that gait adaptation during pregnancy involves coordinated changes across the pelvis–hip complex, knee, and ankle-foot complex, with both kinematic and kinetic alterations reported. Common findings include increased anterior pelvic tilt in late pregnancy, altered hip mechanical demand, phase-specific adjustment of knee flexion during stance, and increased ankle-related mechanical demand during support, shock absorption, and propulsion. However, the consistency of these findings differs across segments: increased anterior pelvic tilt and some hip kinetic changes are reported relatively consistently, knee changes are more often phase-specific, and ankle kinematic findings are less consistent, although kinetic changes appear more robust. These findings suggest that pregnancy-related gait adaptation is not confined to a single joint, but reflects coordinated changes across multiple segments. However, current evidence remains insufficient to distinguish clearly the relative contributions of pregnancy-related morphological change, altered loading, and segmental functional adjustment, or to explain how these changes are mechanically linked within the overall kinetic chain. In particular, without pelvic segment kinetics and pregnancy-specific body parameters, it remains unclear whether the observed joint-level adaptations reflect a systematic pattern of mechanical coordination across the lower limb.

1.3.3 Muscle activation and neuromuscular control during walking in pregnancy

The currently available direct evidence on muscular changes during walking in pregnancy is limited and derives mainly from a small number of studies using surface electromyography. In a longitudinal comparative study, Bagwell et al. recorded kinematics, kinetics, and surface EMG during level walking in 20 pregnant women and 20 nulliparous controls, with bilateral monitoring of the erector spinae, gluteus medius, and gluteus maximus. Compared with nulliparous controls, women in the second trimester showed lower peak gluteus maximus activation during stance. At the same time, the relative contribution of the ankle to negative work was greater, whereas the relative contribution of the hip was smaller, suggesting that changes in gluteus maximus activation occurred alongside altered distribution of mechanical tasks between the hip and ankle [14]. Bagwell et al. later reported an exploratory analysis comparing pregnant women with higher and lower pain scores during pregnancy and postpartum. This study also focused on the erector spinae, gluteus medius, and gluteus maximus, and suggested that pain status may influence both muscle activation and joint-level mechanical task distribution, particularly in relation to reduced relative hip use and increased distal mechanical demand. However, these interpretations remain based primarily on associations between EMG and kinetic findings, and do not yet provide direct evidence of intermuscular coordination mechanisms [50]. Beyond level walking, Gottschall et al. examined muscle activity during transitions between level and inclined walking. In that study, 13 pregnant women were assessed at 20 and 32 weeks of gestation, and surface EMG was recorded from the tibialis anterior, lateral gastrocnemius, biceps femoris, and rectus femoris. As pregnancy progressed, gestational weight gain was reflected by an increase in maternal body mass from 70.88 ± 13.94 kg to 79.46 ± 14.22 kg, and participants showed greater activity in several recorded muscles during hill-related tasks, together with slower walking speed, wider step width, and greater hip, knee, and ankle flexion. The authors interpreted this pattern as a more cautious walking strategy [15].

Existing electromyographic studies suggest that pregnancy-related changes in muscle activity are associated with a redistribution of mechanical demand across the lower limb, particularly during stance, with reduced relative use of proximal hip musculature and increased reliance on distal ankle function. These changes also appear to be influenced by gestational progression, pain status, and task demands, indicating that muscular adaptation is closely related to the evolving requirements for support, balance control, and load transfer during pregnancy gait. However, although current studies provide initial evidence that altered muscle activation accompanies joint-level mechanical reorganization, they remain insufficient to explain in detail how multiple muscle groups coordinate to produce the observed changes in joint motion, segmental acceleration, and overall movement.

1.3.4 Musculoskeletal modeling in pregnancy

During pregnancy, women exhibit a range of notable changes in gait and postural control, including increased lumbar lordosis, anterior displacement of the COM, reduced step length, and adjustments in gait rhythm. While these adaptations may contribute to the maintenance of stability, they are also closely associated with low back pain, muscle fatigue, and an elevated risk of falling. Reliance on traditional kinematic measures and surface-level physiological indicators alone is insufficient to explain why instability arises or to identify which soft tissues and joints are subjected to additional mechanical loading. Consequently, an increasing number of studies have shifted toward internal mechanical analyses based on multisegmental and musculoskeletal modeling. These models represent the human body as an integrated system of bones, joints, and muscles. Joint moments are obtained through inverse kinematics and inverse dynamics, after which muscle forces and joint contact forces are estimated using static or dynamic optimization. This approach enables the reconstruction of pregnancy-specific mechanical loading patterns during walking, quiet standing, and functional transitional movements [52-54].

Systematic reviews indicate that the number of multisegment or pregnancy specific musculoskeletal models developed for pregnant populations remains limited, with only slightly more than a dozen distinct models reported across fewer than twenty studies. Most of these models are three dimensional (3D) and dynamic in nature, with anatomical coverage ranging from the lower extremities to the whole body [55]. These models are predominantly constructed using platforms such as Visual3D, OpenSim, and AnyBody, and rely on marker driven scaling and solution pipelines to estimate joint moments, power, and loading in the pelvic and lumbar regions during tasks including walking, quiet standing, and sit to stand and walking transitions [52-54, 56]. A 3D lower limb and pelvic model representing the second trimester has been used in combination with inverse dynamics to quantify sagittal and frontal plane ankle, knee, and hip moments, revealing a pronounced upregulation of ankle push off and frontal plane hip control during pregnancy [52]. Pregnancy specific rigid body models reconstructed from 3D anthropometric data have further enabled simulations of walking and standing in OpenSim. Compared with generic models, these pregnancy specific representations exert limited effects on ankle, knee, and hip moments but substantially amplify lumbar joint moment demands, providing biomechanical evidence for pregnancy related low back pain and increased fall risk [53].

With respect to model parameterization, most pregnancy specific musculoskeletal models apply pregnancy related revisions to body segment inertial parameters, particularly segment mass, COM location, and moments of inertia in the trunk and pelvic regions, in order to reflect increases in abdominal circumference, segment mass, COM location, and inertial properties associated with gestational weight gain [52-54, 56]. Common approaches include geometric reconstruction based on anthropometric measurements and body shape scanning of pregnant women, as well as the derivation of regression equations from military populations and pregnancy databases to estimate segment mass and COM locations [8, 53]. In parallel, multiple studies have indicated that pregnancy related edema and soft tissue accumulation amplify soft tissue artefact and increase errors in the localization of bony markers, thereby posing significant challenges to traditional linear scaling strategies [52, 55]. In the domain of muscle tendon dynamics, most existing models continue to adopt generic parameters derived from non-pregnant populations, such as optimal muscle fiber length, tendon slack length, and maximal isometric force, with limited calibration based on pregnancy specific measurements of active and passive properties. In addition, static optimization methods have limited capacity to adequately represent co-contraction of muscles in the lumbar and pelvic regions, which may lead to an underestimation of the muscle force and metabolic cost required to maintain stability. To address this limitation, global optimization approaches, including genetic algorithms, have been explored to explicitly model antagonist muscle co-contraction patterns, thereby providing a closer correspondence to clinically observed manifestations of lumbar muscle tension and pain [57].

Overall, existing pregnancy specific musculoskeletal models provide an important bridge between external manifestations, such as reduced step length, increased step width, anterior displacement of the COM, and altered postural sway, and internal mechanisms, including the redistribution of joint moments, increased muscle force demands, and elevated contact forces in the lumbar and pelvic regions. These models also offer a mechanical basis for explaining the increased prevalence of low back pain and fall risk during pregnancy. However, systematic reviews have simultaneously highlighted persistent limitations, including the small number of available models, insufficient validation procedures such as the lack of systematic electromyography verification, the reliance of body segment inertial parameter regressions on non-pregnant populations, and the inadequate representation of complex joint degrees of freedom in the pelvis and lumbar spine. These issues continue to restrict the broader application of current models in clinical and ergonomic contexts [55].

1.4 Walking-related dynamic stability, fall relevance, and footwear as a modifiable factor

1.4.1 Falls during pregnancy and their clinical relevance

Falls during pregnancy represent a clinically relevant safety concern, but they should be understood as multifactorial events rather than as direct consequences of impaired balance alone. Epidemiological studies indicate that fall occurrence is stage-specific and context-dependent, reflecting the interaction between pregnancy-related biomechanical adaptations and behavioral or environmental exposures. For example, a large population-based survey of 3,997 postpartum women in the United States reported that 27% experienced at least one fall during pregnancy, with recurrent falls occurring in over one-third of these cases and approximately one-fifth requiring medical attention [58]. Falls were most frequently reported during late gestation and occurred predominantly in indoor environments, particularly during stair negotiation or on slippery surfaces. Behavioral factors such as hurried walking, failure to use handrails, or wearing unstable footwear further increased fall likelihood.

Similar patterns have been observed in other populations. A multicenter cross-sectional study in Nigeria reported a fall incidence of approximately 25%, with the third trimester representing the period of highest risk [59]. Independent risk factors included maternal age, multiparity, multiple gestation, and unintended pregnancy. Falls most often occurred during household activities, especially when visual obstruction or inappropriate footwear was involved. Although many falls did not result in severe complications, a small proportion were associated with adverse obstetric outcomes, including uterine contractions, vaginal bleeding, or threatened preterm birth [58, 59].

These findings suggest that fall risk during pregnancy emerges from the combined effects of structural adaptations, behavioral strategies, and environmental challenges. Among potentially modifiable external factors, footwear warrants particular attention because it directly mediates the mechanical interaction between the body and the ground. Gestational weight gain-related anterior displacement of the COM, increased pelvic tilt, ligamentous laxity, and alterations in foot morphology may increase the demands placed on locomotor stability, particularly during late gestation [60]. However, the biomechanical pathways through which footwear influences walking stability remain insufficiently characterized. This gap provides an important clinical and research rationale for examining walking-related dynamic stability during pregnancy. Biomechanical relevance should not be equated directly with clinical fall prevention, because actual fall events are also shaped by behavior, environment, and longer-term adaptation.

1.4.2 Current evidence on postural and walking-related stability during pregnancy

Current research on balance and postural stability during pregnancy encompasses a wide range of task paradigms, including quiet standing, device-based dynamic postural assessments, gait-related tasks such as steady walking, gait initiation, turning, and obstacle crossing, as well as more complex functional task sequences, for example rising from a chair, walking, and sitting down again.

Evidence from device-based assessments and standing tasks generally indicates that, as pregnancy progresses, particularly in late pregnancy, the control demands required to maintain balance increase. Inanir et al. used the Biodex Balance System to assess 80 pregnant women across the three trimesters and 30 non-pregnant controls. Their results showed that, at platform stability level 8, the overall, anterior-posterior, and medial-lateral stability indices were all significantly higher in women in late pregnancy than in the control group ($p < 0.05$). In addition, fall risk scores in late pregnancy were significantly higher than those in early and mid-pregnancy [61]. In studies of quiet standing, center-of-pressure measures such as sway path, sway velocity, and sway area are commonly used indicators of stability. A prospective longitudinal study found that, under eyes-open standing conditions, static stability remained largely unchanged during pregnancy and postpartum. Under eyes-closed conditions, however, anterior-posterior sway path length and sway velocity increased in late pregnancy and decreased postpartum ($p = 0.014$ and $p = 0.017$, respectively), suggesting that restriction of visual input may reveal balance control demands that are not apparent under eyes-open conditions [62]. Another recent study involving 19 pregnant women during single-leg standing also reported reduced sway amplitude and sway velocity in mid- and late pregnancy, together with shorter eyes-closed standing time in late pregnancy. This pattern is more likely to reflect the adoption of a more protective or stiffened postural control strategy rather than an actual improvement in balance capacity [63].

Taken together, these studies indicate that pregnancy places greater demands on the postural control system, and that this effect becomes more pronounced in late pregnancy. However, changes in pregnancy-related stability are not uniform across all outcome measures, but are instead influenced by multiple factors, including task type, sensory condition, and individual control strategy. Apparent reductions in sway do not necessarily indicate improved balance performance; rather, they may reflect a protective or stiffened mode of postural control adopted by pregnant women to maintain safety.

1.4.3 Footwear as a modifiable mechanical factor during walking

Footwear is one of the most direct and most readily modifiable external factors influencing gait in daily life. Features such as heel height, heel-toe drop, sole hardness, sole thickness, and upper structure can all alter foot-ground interaction, lower-limb alignment, and the distribution of mechanical loads during the stance and push-off phases. During pregnancy, the importance of footwear may increase further because the foot itself undergoes a series of structural and functional changes. Pregnancy-related foot edema and load adaptation have been shown to be associated with increases in foot length, foot width, and foot volume, reductions in arch height, and alterations in plantar pressure distribution. It has been reported that metatarsophalangeal joint circumference may increase by approximately 9–28 mm in late pregnancy, suggesting that conventional footwear designed for non-pregnant women may not provide optimal fit or sufficient support [34]. In this context, relatively low heel heights appear more appropriate than higher-heeled designs for pregnant women, although the biomechanical effects of specific heel-toe drops remain insufficiently studied [64]. These observations suggest that footwear selection during pregnancy should not be considered solely from the perspective of comfort, but also as an important mechanical factor that may influence walking performance.

Existing research on footwear during pregnancy has focused primarily on comfort, pain relief, plantar pressure redistribution, or orthotic support, while gait-related biomechanical outcomes have received much less direct attention. Previous studies have examined unstable shoes in postpartum women, insoles designed to reduce loading in the first metatarsophalangeal joint region, silicone insoles intended to regulate plantar pressure, and balance shoes aimed at reducing foot loading and promoting circulation [65-69]. Overall, these studies suggest that footwear interventions may alter plantar pressure distribution, foot comfort, and certain balance-related measures. However, the current evidence remains limited, and the reported findings are markedly heterogeneous. The effects described across different study designs and intervention types are not entirely consistent. In addition, many previous studies involved postpartum populations, used insoles rather than whole-shoe interventions, or focused mainly on pain- and comfort-related outcomes. As a result, their direct relevance to walking stability in late pregnancy remains limited.

Therefore, the potential value of “more appropriate footwear” should be defined from a biomechanical perspective rather than assumed in general terms. More appropriate footwear may create conditions that are more favorable for stable walking by improving fit, reducing local discomfort, redistributing plantar loads, and modifying stance-phase mechanics. At the same time, footwear may also shift mechanical loads to other parts of the body. Consequently, improvements in local foot mechanics do not necessarily imply consistent benefits across the entire lower-limb kinetic chain. This issue is particularly important during pregnancy, because postural changes associated with gestational weight gain and the anterior distribution of the added load have already affected whole-body alignment and load transfer. On this basis, the current evidence supports viewing footwear as a modifiable mechanical factor whose effects may be beneficial, neutral, or potentially adverse, depending on the design characteristics of the shoe and the outcome measures under consideration.

Among footwear design parameters, heel-toe drop is of particular importance because it directly influences sagittal-plane alignment and the sequence of plantar loading from initial contact onward. By altering the vertical relationship between the rearfoot and forefoot, heel-toe drop may further affect ankle posture, shank orientation, body center-of-mass position, and the organization of the stance phase. Despite its clear mechanical relevance, its role in pregnancy-related walking biomechanics and stability-related indicators still lacks independent and systematic investigation. Existing studies have mainly focused on insoles, sole structure, or overall comfort, and there is still a clear lack of understanding of how specific footwear structural parameters influence gait biomechanics during pregnancy. In fact, any footwear-induced change in foot position, rollover mechanics, or sagittal-plane alignment may redistribute mechanical demands to more proximal joints. Therefore, the effects of footwear should be evaluated from the perspective of whole-limb, and even whole-gait, biomechanics rather than being limited to local foot effects alone. This gap in the literature provides an important background for the present thesis and constitutes the theoretical rationale for examining heel-toe drop as a testable external design variable within the framework of pregnancy-related walking biomechanics and stability-related indicators.

1.5 Aims and hypotheses

This thesis examined pregnancy-related walking stability from three related perspectives: structural response of the foot, walking biomechanics in late pregnancy, and the short-term biomechanical effects of footwear with different heel-toe drops. Accordingly, the aims of the thesis were organized into three parts.

1.5.1 Structural Adaptation and Foot Arch Response

The first aim of this study was to examine the relative mechanical effects of gestational weight gain-related loading and reduced passive plantar ligament–fascia stiffness on foot arch deformation under simulated loading conditions. Particular attention was given to changes in transverse arch, and the associated redistribution of internal stress within the midfoot. It was hypothesized that increased vertical loading and reduced passive plantar ligament–fascia stiffness would both influence foot arch deformation, but with different mechanical roles.

1.5.2 Walking biomechanics and model-based induced acceleration analysis

The second aim of this study was to characterize walking adaptations during late pregnancy using spatiotemporal gait assessment, three-dimensional gait analysis, force-platform measurements, and pregnancy-adjusted musculoskeletal modeling. Induced acceleration analysis was used to examine IAA-derived muscle-force-induced joint angular acceleration patterns during stance. It was hypothesized that late-pregnancy walking would be characterized by slower walking speed, shorter stride length, lower joint moments, and reduced ankle plantarflexion-related angular acceleration during stance.

1.5.3 Footwear intervention and stability-related biomechanical effects

The third aim of this study was to determine whether footwear with different heel-toe drops could modify walking biomechanics and selected biomechanical indicators associated with stability during late pregnancy. To address this question, gait was compared under three footwear conditions with different sole structures, namely negative-heel shoes (NHS), flat shoes, and low-heel shoes. Particular attention was given to plantar loading patterns, spatiotemporal gait parameters, joint mechanics, and center-of-mass–center-of-pressure relationships during walking. It was hypothesized that, compared with flat shoes and low-heel shoes, NHS would reduce the posterior inclination of the COM–COP vector and alter lower-limb mechanics in a manner consistent with stability-related walking biomechanics. This hypothesis was limited to short-term biomechanical effects observed during walking and was not intended to directly test long-term fall prevention efficacy.

2. Materials and methods

2.1 Participants

2.1.1 Finite element modeling subject

A non-pregnant healthy adult female (n = 1) was selected as the modeling participant for the finite element analysis component of this study. The participant was 25 years of age, with a height of 163 cm and a body mass of 57.3 kg. The participant exhibited a normal foot arch structure and had no history of musculoskeletal disorders. All study procedures were approved by the Medical Ethics Committee of Ningbo University (TY2025070) and conducted in accordance with the Declaration of Helsinki; written informed consent was provided by the participant. This finite element model was constructed from the CT data of a single participant and therefore represents a subject-specific foot geometry rather than a population-based anatomical model.

2.1.2 Gait comparison study participants

To investigate gait adaptations and biomechanical interpretation based on musculoskeletal modeling strategies during pregnancy, a comparative study design was adopted. Participants were recruited in accordance with previously reported inclusion criteria [14]. A total of 40 women were enrolled and divided into two groups: a pregnant group (N = 20) representing late gestation and a non-pregnant control group (CON, N = 20). Written informed consent was obtained prior to participation, and the study was approved by the Ethics Committee of Ningbo University (RAGH202201154396.6). Inclusion criteria for the pregnant group included an age range of 25–35 years and a gestational age of approximately 32 weeks. Exclusion criteria comprised a history of lower-limb surgery, neurological disorders, or musculoskeletal injuries that could affect gait performance. The control group consisted of healthy non-pregnant women with no history of lower-limb pathology. The anthropometric characteristics of both groups are presented in Table 2.1. The pregnant group had a mean gestational age of 32.25 ± 5.36 weeks, waist circumference of 0.98 ± 0.07 m, age of 29.00 ± 3.94 years, body mass of 67.00 ± 7.51 kg, height of 1.66 ± 0.05 m, and a BMI of 25.30 ± 3.68 kg/m². The control group exhibited comparable age and height but lower body mass and BMI. All participants provided written informed consent after receiving a detailed explanation of the experimental procedures. This cohort configuration was established to support induced acceleration analysis, enabling the examination of pregnancy-related differences in model-estimated muscle-induced joint angular accelerations during gait.

Table 2.1. Basic characteristics of the participants for the gait comparison study.

	CON (N=20)	Pregnant women (N=20)
Gestational age (weeks)	/	32.25 ± 5.36
Age (yr)	27.00 ± 4.58	29.00 ± 3.94
Weight (kg)	56.10 ± 4.39	67.00 ± 7.51
Height (m)	1.65 ± 0.05	1.66 ± 0.05
BMI (kg/m²)	20.56 ± 1.35	25.30 ± 3.68

Note: CON, Control group.

2.1.3 Footwear intervention participants

To examine whether footwear structure could modulate walking biomechanics in late pregnancy, a controlled repeated-measures study was designed. An a priori sample size calculation was conducted using G*Power 3.1 for a repeated-measures ANOVA with three footwear conditions [70]. A moderate expected effect size of f = 0.30, $\alpha = 0.05$, power = 0.80, an assumed correlation among repeated measures of 0.50, and $\varepsilon = 1.00$ were used. This calculation indicated a minimum required sample size of 20 participants. To account for potential dropout, unusable gait trials, and incomplete motion-capture or force-platform data, and consistent with previous pregnancy-related foot/orthosis research allowing for attrition [67], 28 women in late pregnancy were recruited.

The anthropometric characteristics of the participants are summarized in Table 2.2. Strict inclusion and exclusion criteria were applied to isolate the biomechanical effects of footwear from other pathological confounders. Inclusion criteria required a gestational age of at least 29 weeks, no history of lower-limb surgery, and the absence of serious pregnancy-related complications. To prevent structural abnormalities from obscuring the intervention effects, exclusion criteria comprised foot deformities, marked musculoskeletal pain, pre-existing balance impairments, or severe foot edema. Participants completed gait assessments under three randomized conditions: negative-heel shoes, flat shoes, and low-heel shoes. This design allowed for a direct comparison of how varying heel-toe drops influence the COM trajectory and joint kinetics. Written informed consent was obtained prior to participation, and the study was approved by the Ethics Committee of Ningbo University (RAGH202201154396.6).

Table 2.2. Anthropometric characteristics of the recruited participants (N = 28).

Information	**Mean**	**S.D.**
Age (year)	28.4	2.30
Height (m)	1.63	0.04
Gestational age (w)	33.43	3.37
BMI (kg/m2)	26.6	3.52

2.2 Experimental protocol and data collection

2.2.1 Finite Element Modeling Procedures

Computed tomography (CT) imaging of the foot was performed with the ankle joint positioned in neutral alignment. In addition, biplanar standing radiographs were acquired and used to register the reconstructed foot model to the physiological weight-bearing posture (Figure 2.1A). A total of 1079 axial slices were acquired, with a slice thickness of 0.5 mm and a reconstruction interval of 0.25 mm. These images were used to construct a complete 3D foot model incorporating skeletal structures, articular cartilage, surrounding soft tissue envelopes, and ligamentous components. To simulate the biomechanical effects of gestational weight gain-related loading and parametrically reduced passive plantar ligament–fascia stiffness on foot structure and transverse-arch mechanics, a three-dimensional (3D) finite element model was constructed. A total of 28 foot bones and surrounding soft tissues were reconstructed based on CT data obtained (Figure 2.1B). Image segmentation was performed using a gray-value thresholding method in Mimics (v.21.0, Materialise, Leuven, Belgium). The segmented surfaces were subsequently smoothed in Geomagic Studio (v.2024, 3D Systems, Rock Hill, SC, USA) to remove local artefacts and improve geometric continuity. The reconstructed structures were then assembled into a complete three-dimensional foot model in SolidWorks (v.2021, Dassault Systèmes, Vélizy-Villacoublay, France). Finite element mesh generation was subsequently performed in ANSYS Workbench 2021 R1 (ANSYS Inc., Canonsburg, PA, USA) using four-node tetrahedral elements. Mesh convergence was evaluated by progressively reducing element size until the change in peak plantar contact pressure was less than 5%, ensuring numerical solution stability.

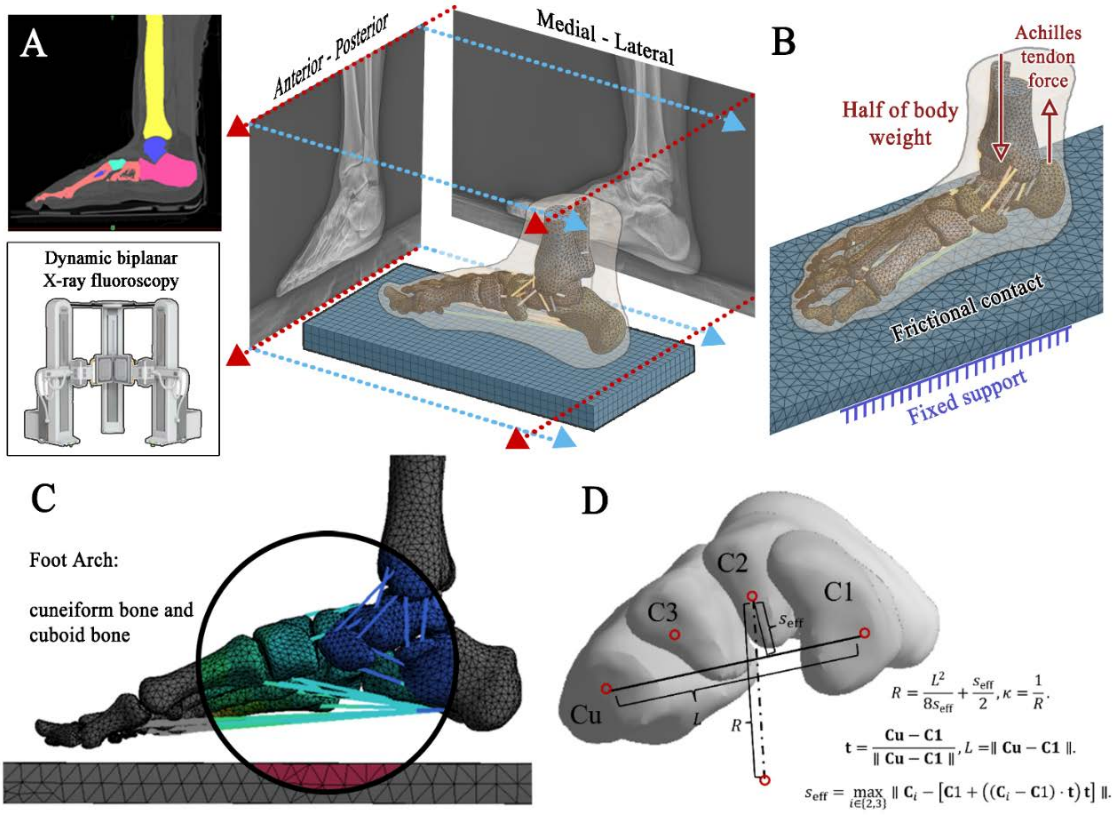


Figure 2.1. Workflow for finite element (FE) model development and geometrical definition of the tarsal transverse arch.

(A) CT segmentation, 3D reconstruction, biplanar standing X-ray registration, and FE model construction. (B) Generation of the full skeletal model, including mesh construction and assembly of bone components. (C) Finite element model incorporating bones, ligaments, cartilage layers, and the plantar fascia. (D) Geometrical definition of the tarsal transverse arch and calculation of structural indicators. The arch is defined using anatomical landmarks on the medial (C1), intermediate (C2), and lateral (C3) cuneiforms and the cuboid (Cu). Chord length L spans the cuneiforms, effective rise height S_eff is measured orthogonally from the chord to the apex, and curvature is derived from the radius R fitted to the osseous arc.

To assess the physiological plausibility of the FE model, the plantar pressure distribution predicted by the model was compared with in-shoe plantar pressure data obtained using the Pedar in-shoe system (Novel GmbH, Munich, Germany) from the same participant during static standing. Peak plantar pressure in the forefoot and rearfoot was used because these regions represent the principal load-bearing zones during quiet standing. The forefoot-to-heel peak pressure ratio was examined to characterise anterior–posterior load sharing. To provide a reference-condition external check for the FE model, the forefoot and heel peak plantar contact pressures predicted under the non-pregnant reference condition were compared with Pedar in-shoe pressure data obtained from the same participant during static standing. Regional peak-value differences were calculated within the forefoot and heel regions. This comparison was limited to the reference condition and was interpreted as an external plausibility check of regional plantar contact pressure.

A total of 28 bones were reconstructed, and 51 major foot ligaments were modeled based on anatomical data, with attachment points refined according to the CT geometry [71-73]. Articular cartilage and connective tissues were represented by solid layers with a thickness of 2 mm inserted between articular bones. The plantar fascia was constructed using truss elements extending from the medial calcaneal tubercle to the proximal phalanges.

Four physiological conditions were defined on a single foot finite element geometry using longitudinal data on body weight and soft tissue mechanics spanning the non-pregnant state to late pregnancy. These conditions comprised the non-pregnant baseline condition and gestational ages of 9.1, 20.9, and 32.3 weeks, and shared identical bone morphology and mesh discretisation. Pregnancy-related biomechanical variation was parameterized by two factors, the single-foot body weight (W) and the stiffness of the plantar ligaments and the plantar fascia (K). Body mass increased from 57.3 kg in the non-pregnant state to 65.3, 68.1, and 73.5 kg during pregnancy [23], corresponding to single-foot vertical loads of 281, 320, 334, and 361 N respectively under symmetric bilateral stance. These loads were applied vertically to the superior surface of the talus to represent gravitational loading during standing. To represent pregnancy-related increases in passive soft-tissue compliance, the Young's modulus of the plantar ligaments and plantar fascia was scaled by 5.3%, 15.2%, and 14.4% across the three gestational stages. These reductions were implemented as relative stiffness-scaling factors rather than as direct measurements of plantar-tissue material properties, and were derived from published data describing changes in peripheral joint laxity during pregnancy. Unless otherwise specified, all tissues were modeled as homogeneous, isotropic, and linearly elastic. Material properties were assigned based on previously published experimental and computational studies. Cortical bone was assigned a Young's modulus (E) of 17,000 MPa and a Poisson's ratio (ν) of 0.3, consistent with its predominantly cortical composition [74, 75]. Articular cartilage was defined with $E = 1$ MPa and $\nu = 0.4$ [76]. Encapsulated soft tissue was prescribed a low stiffness ($E = 0.15$ MPa) and a high Poisson's ratio ($\nu = 0.45$) to approximate its compliant and nearly incompressible behavior [77].

Ligaments were represented using two node truss elements that could bear tension but not compression, which prevented non physiological resistance in the compressive direction. Nine primary ligaments that mainly sustain load were modeled with hyperelastic material properties. These ligaments included the anterior talofibular, anterior tibiofibular, calcaneofibular, posterior talofibular, tibionavicular, spring, tibiocalcaneal, anterior tibiotalar and posterior tibiotalar ligaments. Their cross sectional areas ranged from 1.5 to 5.91 mm², based on anatomical measurements [78]. Secondary ligaments that mainly contribute to stability, including the cervical ligament, the talonavicular ligament, the interosseous membrane and the posterior interosseous ligament, were modeled as linearly elastic truss elements with $E = 264.8$ MPa and a cross sectional area of 1.784 mm² [74, 75]. All ligament elements were assigned a uniform density of 1,000 kg/m³. The plantar fascia was represented using hyperelastic truss elements that could carry tension but not compression, spanning from the medial calcaneal tubercle to the proximal phalanges. Its stiffness parameters were calibrated within the reported effective range of 0 to 700 MPa [74, 75, 79]. The plantar surface interacted with a rigid floor, which was defined with $E = 14{,}200$ MPa, $\nu = 0.1$, and a friction coefficient of 0.5. A complete summary of all material models and parameters is provided in Table 2.3. The rigid floor was fully fixed in all translational and rotational degrees of freedom. The vertical load applied to the talus was constrained to act only in the vertical direction. These constraints were introduced to prevent rigid-body motion while allowing physiologically relevant deformation of the foot arches (Figure 2.1C).

Table 2.3. Material properties used in the FE model.

Component	Element Type	Young's modulus (MPa)	Density (kg/m³)	Poisson's Ratio ν	Element Size (mm)
Bone	3D tetrahedral	17,000	1990	0.3	3.24
Cartilage	3D tetrahedra	1	N/A	0.4	0.81
Soft tissue	3D tetrahedral	0.15	950	0.45	3.24
Floor	3D tetrahedral	14,200	1000	0.1	8.10
Primary ligaments	2-node truss (tension only)	Hyperelastic; Area: 1.5–5.91 mm²	1000	N/A	N/A
Other ligaments	2-node truss (tension only)	264.8, Linear elastic; Area: 1.784 mm²	1000	0.4	N/A
Plantar fascia	2-node truss (tension only)	0–700; Hyperelastic; Area: 290.7 mm²	1000	N/A	N/A

2.2.2 Gait Comparison Protocol

The experimental protocol of the present study comprised wearable gait monitoring, 3D motion capture assessment, and synchronized acquisition of ground reaction force data. This integrated approach was designed to comprehensively obtain spatiotemporal gait parameters, as well as kinematic and kinetic data.

Spatiotemporal gait characteristics were collected using the IDEEA 3 wearable gait analysis system (MiniSun LLC, Fresno, CA, USA), which sampled acceleration signals at 64 Hz. The IDEEA system has previously shown good concurrent validity for spatiotemporal gait parameters (ICC = 0.784–0.998), although walking speed, step length, and stride length may be underestimated by about 7% relative to force-platform measurements. Its intrasession reliability has also been reported to be high (CV < 5.7; ICC > 0.961) [80]. This system consists of a primary recording unit worn at the waist and subsidiary recording units positioned at the ankles. Data acquisition was achieved using five sensors mounted on the sternum, the mid-thigh, and the bilateral fourth metatarsophalangeal joints, enabling synchronous recording of three-dimensional acceleration and angular changes of multiple body segments. The software interface of the system is illustrated in Figure 2.2.

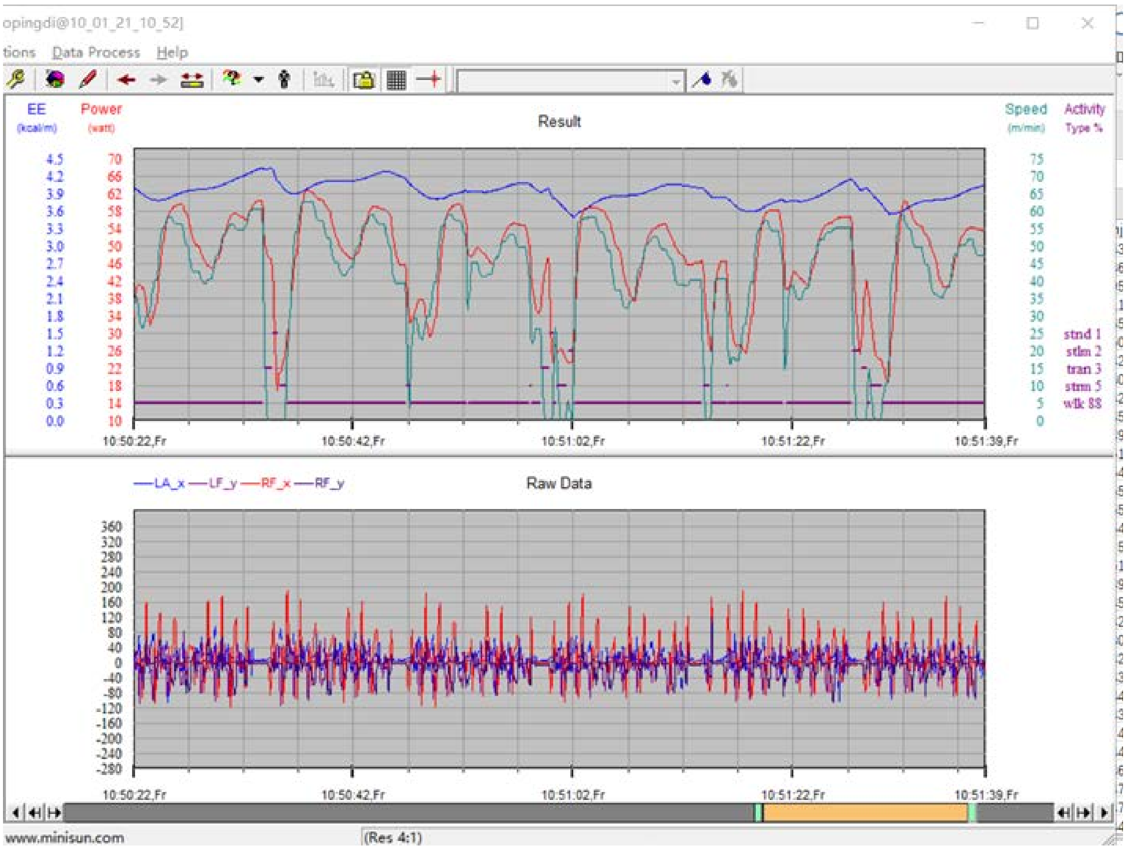


Figure 2.2. Software interface for gait-related data processing and visualization, including energy expenditure, power output, walking speed, and raw sensor signals.

After the devices had been fitted, participants first walked freely in a flat and open area to become familiar with the equipment and the environment. During the formal trials, participants completed the walking task at a self-selected comfortable speed along the predefined route illustrated in Figure 2.3. The system automatically recorded gait speed, cadence, step length, thigh acceleration, mechanical work during the swing phase, impact intensity at initial contact, and the proportions of single- and double-support time. For the wearable assessment, participants completed a 50 m overground walking task, from which spatiotemporal gait parameters were later extracted.

Figure 2.3. Laboratory walkway used for the 50 m free-walking gait assessment.

To minimize data bias arising from external interference, all participants were provided with the same model of athletic shoes and socks, and the entire testing area was uniformly covered with a rigid wooden floor. Under laboratory conditions, kinematic and kinetic data were collected using a Vicon motion capture system with eight infrared cameras (Vicon Motion Systems Ltd., Oxford, UK). The system comprised eight infrared high-speed cameras operating at 200 Hz and one AMTI force platform (Advanced Mechanical Technology, Inc., Watertown, MA, USA) three-dimensional force platform sampling at 1000 Hz. Prior to testing, 46 retroreflective spherical markers (diameter: 12 mm) were attached to anatomical landmarks, with marker locations illustrated in Figure 2.4. The camera system recorded three-dimensional marker trajectories during walking, while the force platform synchronously measured ground reaction forces in three directions during foot contact with the ground. Participants walked across the force platform using a self-selected comfortable speed gait pattern and were not given any instructions regarding step length or deliberate foot placement. A 3-min rest interval was provided between walking trials to minimize participant fatigue. Only the dominant lower limb was analyzed. The dominant limb was defined as the preferred kicking leg, so that the same functional side was compared across participants and conditions.

Figure 2.4. Experimental setup for gait data collection in pregnant participants using a motion capture system.

In the present study, induced acceleration analysis (IAA) was used as a model-based biomechanical inference approach to quantify the model-predicted effects of selected muscle groups on segmental and joint accelerations during walking. In this approach, the contribution of each muscle group is inferred from the acceleration it induces in the musculoskeletal model through its muscle-generated force or joint moment, while accounting for the dynamic coupling among body segments. Therefore, IAA outputs were interpreted as indirect indicators of the mechanical roles of the selected muscle groups during gait, rather than as direct measurements of muscle force or muscle activation. The muscle groups included in the IAA were selected a priori on the basis of their established functional relevance to pregnancy-related gait adaptation and stance-phase stability. The gluteus, iliopsoas, hamstrings, quadriceps, soleus, gastrocnemius, and tibialis anterior were analyzed because these muscle groups are primary contributors to proximal support, limb progression, shock absorption, and ankle push-off during walking, and have been consistently implicated in previous gait biomechanics studies of pregnancy and lower-limb stability.

2.2.3 Footwear Intervention Protocol

For the footwear intervention experiment, each participant completed the walking assessment procedures under three footwear conditions: negative-heel shoes, flat shoes, and low-heel shoes. The order of the footwear conditions was randomized before testing. Prior to formal data collection, participants were familiarized with the testing environment and the experimental footwear. Because the negative-heel shoes differed substantially from habitual footwear, a 10-min acclimatization walking period was provided under this condition to minimize unfamiliarity-related effects.

Under each footwear condition, participants completed both a 50 m free-walking assessment and a laboratory-based gait capture test at a self-selected comfortable walking speed. No external pacing cues were provided. In the 50 m free-walking assessment, participants walked naturally after receiving the start command. Each footwear condition was tested three times, with a 3-min rest interval between repeated trials. After completing one footwear condition, participants rested for approximately 5 min before changing to the next randomized footwear condition. For the laboratory-based gait capture test, participants walked along a 10 m walkway while maintaining a natural and stable gait pattern, looking forward with their arms swinging naturally at their sides. A valid trial was defined as one in which the dominant foot contacted the AMTI force platform naturally without obvious targeting, and complete kinematic and kinetic data were recorded. Three valid trials were collected for each footwear condition, and the mean value was used for subsequent analysis. Throughout the testing procedure, participant discomfort, pain, or abnormal gait responses were monitored by the research team together with obstetric clinicians to ensure participant safety.

Three categories of experimental footwear characterized by distinct heel-toe drops were employed in this study: negative-heel shoes, flat shoes, and low-heel shoes. To ensure consistency, all footwear featured identical canvas uppers and rubber soles, with sizes ranging from 36 to 40. The negative-heel shoes had a heel height of 10 mm and a forefoot height of 25 mm, resulting in a −15 mm heel-toe drop. The flat shoes maintained a uniform height of 10 mm at both the heel and forefoot, corresponding to a 0 mm drop. The low-heel shoes had a heel height of 25 mm and a forefoot height of 10 mm, producing a +15 mm drop (Figure 2.5).

Figure 2.5. Experimental footwear conditions used in this study: negative-heel shoes, flat shoes, and low-heel shoes (from left to right).

2.3 Data processing

2.3.1 Finite Element Data Processing

Key geometrical and mechanical indicators of the transverse arch were extracted from the finite element results to characterize its responses under different W–K combinations according to the structural definition shown in Figure 2.1D. Geometrical indicators were computed according to the structural definition of the transverse arch [81, 82], including chord length, effective rise height, curvature, and the spatial position of the arch apex. Curvature was calculated as the reciprocal of the radius fitted to the osseous arc of the transverse arch. Normalized curvature was expressed as $\kappa/\kappa_{ref} \times 100\%$, where κ_{ref} represents the curvature under the nonpregnant reference condition. Mechanical indicators included peak von Mises stress within the transverse arch region and plantar pressure distribution.

To systematically evaluate the main effects of W and K and their interaction, W and K were each divided into four levels within their physiological ranges, yielding a full-factorial design with sixteen simulation conditions. The four pregnancy related conditions corresponded to specific points within this parameter space. All simulations used identical geometry, material definitions, and boundary conditions, ensuring that model responses were driven solely by variations in W and K [83].

2.3.2 Gait Data Processing

Given the substantial pregnancy-related alterations in body weight and segmental mass distribution relative to generic human models, a pregnancy-specific OpenSim 4.1 (Stanford University, Stanford, CA, USA) musculoskeletal model was employed [53]. Segmental mass parameters and COM locations within the model were adjusted to account for pregnancy-related anthropometric changes. Marker trajectories and ground reaction force data were formatted using a Matlab (v. R2016b, MathWorks, Natick, MA, USA) script. Prior to musculoskeletal analysis, kinematic and force data were low-pass filtered using zero-lag fourth-order Butterworth filters with cutoff frequencies of 12 Hz and 30 Hz, respectively. Subsequently, the weights and positions of virtual markers on body segments were manually optimized based on initial scaling results to maximize concordance with experimental marker coordinates. This scaling procedure ensured anthropometric consistency, yielding a root mean square error (RMSE) between experimental and virtual markers of less than 0.02 m, with a maximum error below 0.04 m [84-86]. Joint angles were computed using the Inverse Kinematics algorithm, which solves a weighted least squares problem to minimize marker tracking errors. For the wearable assessment, participants completed a 50 m overground walking task, from which spatiotemporal gait parameters and segmental acceleration variables were later extracted.

This wearable assessment was used to capture gait characteristics over multiple consecutive steps during natural overground walking, whereas the laboratory-based motion capture system was used for joint-level kinematic, kinetic, and OpenSim analyses.

Induced Acceleration Analysis (IAA) was performed using the OpenSim analysis tool as a model-based method for estimating muscle-force-induced joint angular accelerations during the stance phase of walking. The IAA-derived muscle-force-induced joint angular accelerations were extracted and reported in deg/s². The signs followed the joint-coordinate conventions of the musculoskeletal model. Positive values represented hip flexion, knee extension, and ankle dorsiflexion, whereas negative values represented hip extension, knee flexion, and ankle plantarflexion. Accordingly, the results were interpreted according to both the direction and the magnitude of the induced angular acceleration. In this framework, the IAA-derived value represents the joint angular acceleration induced within the musculoskeletal model by the force assigned to a selected muscle group. This distinction arises because joints are subject to simultaneous forces that may superimpose or cancel each other out [87, 88]. This relationship is governed by the system's equations of motion:

$$\ddot{q}_{total} = \sum_{i=1}^{n} \ddot{q}_i^{mus} + \ddot{q}^{grav} + \ddot{q}^{vel}$$

In this equation, $\ddot{q}_i^{mus}$ denotes the acceleration induced by the $i-th$ muscle, $\ddot{q}^{grav}$ is the acceleration induced by gravity, and $\ddot{q}^{vel}$ represents accelerations due to velocity-dependent forces [88, 89].

Regarding the kinematic constraints within the simulation framework, OpenSim supports various constraint types for IAA, including point, weld, and rolling surface constraints. A point constraint enforces coincidence between points on two separate bodies while permitting free relative rotation. In contrast, a weld constraint additionally fixes the relative orientation of the bodies. In the current study, the foot-ground interaction was modeled using a no-slip rolling constraint. This approach provides a more physiological representation of the stance phase than a simple hinge joint [90-92]. To validate the accuracy of the decomposition, the superposition principle was applied to the whole-body COM acceleration. The theoretical COM acceleration (a_{CoM}^{calc}) was calculated as follows:

$$\vec{a}_{COM}^{calc} = \sum_{i=1}^{n} \vec{a}_{COM,i}^{mus} + \vec{a}_{COM}^{grav} + \vec{a}_{COM}^{vel}$$

This calculated summation was then compared against the experimentally measured COM acceleration derived from ground reaction forces to assess the dynamic consistency of the simulation [90, 91].

Gait events were identified from the vertical ground reaction force signal. Initial contact was defined as the instant when the vertical ground reaction force first exceeded 20 N, and toe-off was defined as the instant when it fell below 20 N. The stance phase was defined as the period between initial contact and toe-off and was time-normalized to 100% for subsequent analysis.

For each participant included in the IAA analysis, the signed joint angular acceleration induced by each selected muscle group was averaged across the time-normalized stance phase. These participant-level stance-averaged values were then used to calculate group means and standard deviations for Figure 3.7. Because the acceleration could change direction during stance, positive and negative values occurring at different time points could partially cancel. Therefore, the reported values represent the net stance-averaged mechanical effect of each modeled muscle group and should not be interpreted as phase-independent muscle actions. OpenSim was utilized to compute peak joint angles, range of motion (ROM), peak joint moments, muscle forces, and induced accelerations (IA) during the stance phase of gait.
To provide an energy-based description of lower-limb joint function, sagittal-plane joint work was additionally calculated for the hip, knee, and ankle during the stance phase. Joint work was computed as the integral of body-mass-normalized joint moment with respect to joint angular displacement, using $W = \int M \, d\theta$, where M is the joint moment and θ is the joint angle expressed in radians. Positive work, negative work, and total absolute work were calculated separately for each joint. Negative work was reported as the absolute magnitude of the negative portion of joint work. Because joint moments were body-mass normalized, joint work was expressed in J/kg. The analysis focused on primary muscle groups identified as potential contributors to gait alterations, including the gluteus, iliopsoas, hamstrings, quadriceps, soleus, gastrocnemius, and tibialis anterior [88]. These muscles were selected because they represent the principal proximal and distal contributors to stance-phase support, limb advancement, braking control, and push-off mechanics, which were central to the research question of pregnancy-related gait adaptation. As noted previously, the IA calculated via IAA is distinct from the net joint acceleration because opposing moments may cancel each other out [87]. Consequently, the actual joint acceleration was calculated as the summation of all accelerations induced by both muscle and non-muscle factors. No electromyographic (EMG) measurements were collected in the present study. Therefore, the muscle-induced joint angular accelerations estimated by IAA should be interpreted as model-based biomechanical inferences rather than direct validation of neuromuscular activation patterns.

2.3.3 Footwear Data Processing

Gait data collected under the three footwear conditions were processed for the dominant limb. Reflective marker trajectories and ground reaction force signals were filtered using zero-lag fourth-order Butterworth low-pass filters, with cutoff frequencies of 12 Hz for kinematic data and 30 Hz for force data. Initial contact and toe-off were identified using a vertical ground reaction force threshold of 20 N. The stance phase was defined from initial contact to toe-off and was time-normalized to 101 data points using linear interpolation. Ground reaction force components were normalized to body weight. Ground reaction force components in the mediolateral, anteroposterior, and vertical directions were analyzed as stance-phase waveforms. For COP analysis, the mediolateral and anteroposterior COP components were combined as a two-dimensional trajectory, COP(t) = [COPx(t), COPy(t)]. The two-dimensional COP trajectory was plotted using COPx and COPy over the normalized stance phase. The COM–COP inclination angle was calculated as the angle between the COM–COP vector and the vertical reference line passing through the COP. Spatiotemporal variables included walking speed, cadence, stride length, and single-support time. Knee and ankle angles, sagittal-plane knee and ankle moments, and three-dimensional ground reaction force components were extracted during stance. Values from three valid trials were averaged for each footwear condition before statistical analysis.

2.4 Statistical analysis

2.4.1 Finite element model analysis

To describe the relative sensitivity of transverse-arch geometry to body weight-related loading (W), passive plantar ligament–fascia stiffness (K), and their interaction, a standardized regression-based sensitivity analysis was performed using the model-derived outputs from the 16 full-factorial simulation conditions. Body weight was represented by the single-foot vertical load (W), while passive connective-tissue stiffness was represented by the passive plantar ligament–fascia stiffness-scaling parameter (K). Because W and K were defined a priori within an orthogonal full-factorial design and the outputs were deterministic responses from a single subject-specific finite element model, the analysis was used as a descriptive parametric framework rather than as population-level statistical inference. The standardized coefficients were used to compare the direction and relative magnitude of the associations between W, K, their interaction, and the selected transverse-arch geometric indicators. These coefficients were visualized using forest plots to illustrate whether each indicator was more closely associated with increased loading, reduced passive plantar ligament–fascia stiffness, or their combined effect within the simulated design space.

2.4.2 Gait Analysis for Pregnant and Non-Pregnant Groups

Data processing was performed using IDEEA software. Subsequently, statistical analyses were conducted using the SPSS software package (Version 26.0, SPSS Inc., Chicago, IL, USA). The normality of the data distribution was verified using the Shapiro–Wilk test. For gait parameters, between-group differences were examined using independent-samples t-tests when the normality assumption was satisfied and Mann–Whitney U tests when normality was not satisfied. For the 21 stance-averaged IAA-derived muscle-force-induced joint angular acceleration variables, comprising seven muscle groups across the hip, knee, and ankle joints, between-group differences were assessed using Welch's independent-samples t-tests because equal variances were not assumed. Bonferroni-adjusted p values were calculated across all 21 IAA comparisons. An adjusted p value below 0.05 was considered statistically significant, equivalent to an unadjusted significance threshold of $p < 0.0024$. The level of statistical significance was set at $\alpha = 0.05$. For the seven wearable-derived gait parameters reported in Table 3.2, Bonferroni correction was applied across the seven between-group comparisons, with the corrected significance threshold set at $p < 0.0071$. For normally distributed variables, effect sizes were calculated using Hedges' g. For sagittal-plane joint work variables, between-group differences in hip, knee, and ankle positive work, negative work, and total absolute work were analyzed using independent-samples t-tests when normally distributed, or Mann–Whitney U tests when normality was not satisfied. Bonferroni correction was applied across the nine joint-work comparisons, with the corrected significance threshold set at p <

0.0056. For the 15 joint-angle, ROM, and joint-moment variables reported in Table 3.3, Bonferroni correction was applied across the 15 between-group comparisons, with the corrected significance threshold set at $p < 0.0033$.

2.4.3 Gait experiment data analysis

All footwear-experiment data were analyzed using SPSS version 26.0 and MATLAB (v.R2017a, MathWorks, Natick, MA, USA). Continuous variables are presented as mean ± standard deviation. The significance level for omnibus tests was set at $\alpha = 0.05$.

Time-series waveform variables, including knee and ankle joint angles, knee and ankle joint moments, and the three ground reaction force components, were analyzed using one-dimensional Statistical Parametric Mapping (SPM) implemented in MATLAB. Stance-phase waveforms were time-normalized to 101 data points. SPM inference was used to control the familywise error rate across the normalized stance-phase continuum using random field theory. When post-hoc pairwise SPM comparisons were performed among the three footwear conditions, Bonferroni correction was applied across the three pairwise comparisons, with the corrected significance threshold set at $p < 0.017$.

Because COP represents a two-dimensional trajectory, the mediolateral and anteroposterior COP components were analyzed jointly as a two-component time-continuous vector field using vector-field SPM based on Hotelling's T^2 statistics. For each pairwise footwear comparison, paired COP difference vector fields were calculated as $\Delta COP(t) = [\Delta COPx(t), \Delta COPy(t)]$ and tested against the zero vector using SPM (T^2). Bonferroni correction was applied across the three post-hoc footwear comparisons, with the corrected significance threshold set at $p < 0.017$. Discrete variables, including walking speed, cadence, stride length, single-support time, peak COM–COP inclination angle, peak COM–COP forward tilt angle, and peak COM–COP backward tilt angle, were analyzed using repeated-measures statistics.. Normality was assessed using the Shapiro–Wilk test. For normally distributed variables, footwear condition was treated as a within-subject factor and tested using one-way repeated-measures ANOVA. When sphericity was violated, the Greenhouse–Geisser correction was applied. For variables that did not meet normality assumptions, the Friedman test was used. When post-hoc pairwise comparisons were performed among the three footwear conditions, Bonferroni correction was applied, with the corrected significance threshold set at $p < 0.017$.

3. Results

3.1 Finite element analysis of plantar loading and transverse-arch deformation

3.1.1 Reference-condition plantar pressure comparison and FE-simulated plantar contact pressure

Under the nonpregnant reference condition, the model-predicted peak contact pressures within the forefoot and heel regions were 93.71 kPa and 103.12 kPa, respectively. The corresponding Pedar-measured peak pressures within the same regions were 95.13 kPa and 106.24 kPa, giving regional peak-value differences of 1.49% and 2.94%, respectively (Figure 3.1A). These results indicate that the FE model produced forefoot and heel peak pressure magnitudes comparable to the Pedar measurements under the reference static standing condition.

During static standing, peak plantar contact pressure was higher as additional mass was added in three simulation conditions from the nonpregnant reference condition to late pregnancy (Figure 3.1B). Compared with the reference condition, forefoot peak plantar contact pressure increased by 14.2%, 17.2%, and 33.6% in the first, second, and third trimesters, respectively. Heel peak plantar contact pressure increased by 15.6% in the first trimester, showed only a small increase of 3.1% in the second trimester, and then increased by 35.3% in the third trimester. Accordingly, the forefoot-to-heel peak pressure ratio was 0.91 at the reference condition, 0.90 in the first-trimester condition, 1.03 in the second-trimester condition, and 0.90 in the third-trimester condition, indicating a transient relative increase in forefoot loading under the second-trimester W–K input condition.

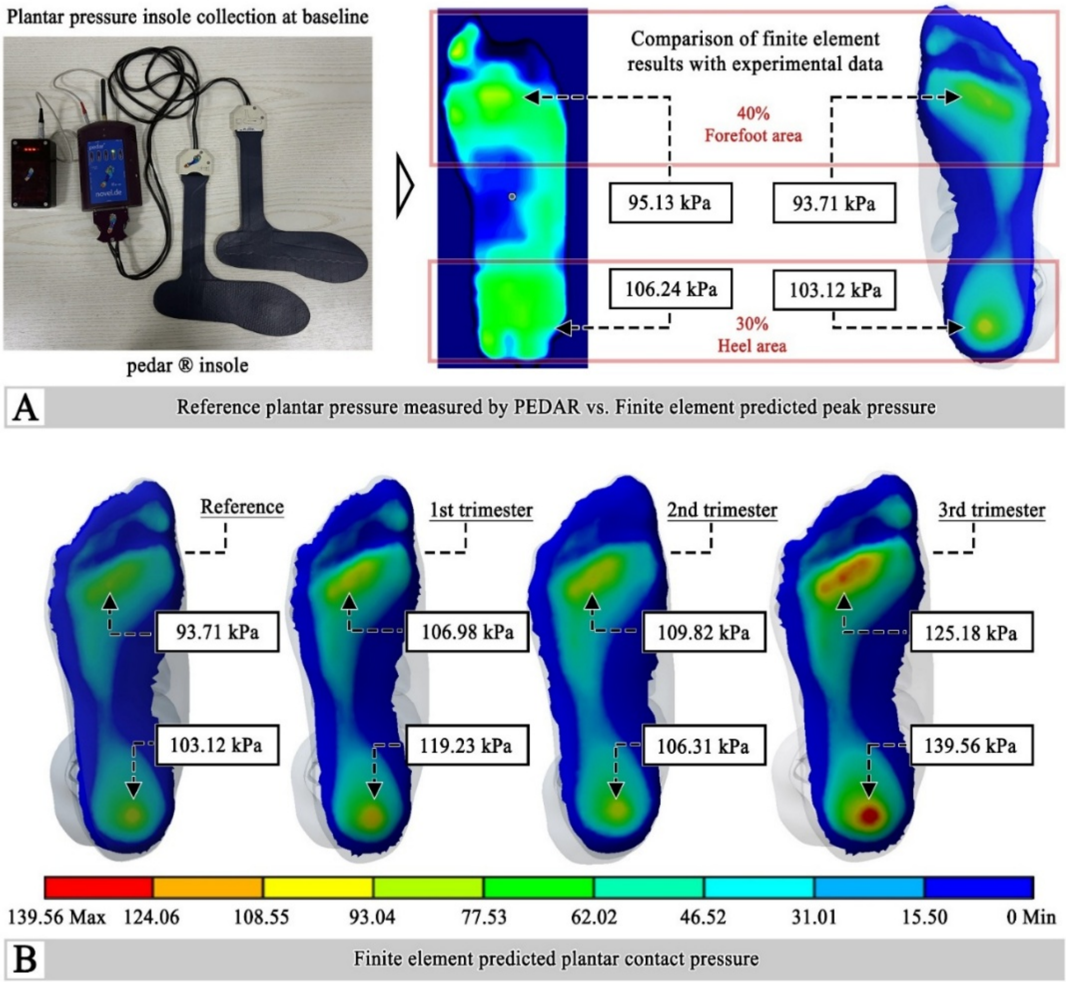


Figure 3.1. Reference-condition plantar pressure comparison and plantar contact pressure simulations.

(A) Pedar measurements and finite element predicted peak pressures in the forefoot and heel regions in the nonpregnant reference condition. (B) Finite element predicted plantar contact pressure for the reference condition and for 9.1, 20.9, and 32.3 weeks of gestation.

3.1.2 Peak von Mises stress in the transverse arch bones

Peak von Mises stress in the transverse-arch bones showed a non-linear pattern across the simulated pregnancy-stage conditions. The reported values represent the peak von Mises stress extracted separately from each transverse-arch bone, including the medial cuneiform (C1), intermediate cuneiform (C2), lateral cuneiform (C3), and cuboid (Cu). The spatial distribution of the stress regions is shown in Figure 3.2B. Peak von Mises stress in the transverse arch bones demonstrated a non-linear pattern across pregnancy. In the medial cuneiform (C1), peak stress increased from 1.36 MPa under the reference condition to 2.78 MPa in the second trimester, representing an increase of approximately 104%, before declining to 1.77 MPa in the third trimester. A similar but less pronounced pattern was observed in the intermediate cuneiform (C2), where stress rose from 0.94 MPa to 1.59 MPa (≈69% increase) and showed a slight reduction thereafter. In contrast, the lateral cuneiform (C3) exhibited consistently low stress levels with only minor variation across conditions (0.41–0.58 MPa). Unlike C1, C2, and Cu, C3 reached its maximum stress in the third trimester, although the absolute stress level remained low, indicating a smaller model-predicted stress response in C3 under the present loading and stiffness conditions. The cuboid (Cu) also showed a pronounced stress response. Peak stress increased from 1.23 MPa in the reference condition to 2.48 MPa in the second trimester (approximately 102%), followed by a reduction to 1.61 MPa in the third trimester. Thus, the largest numerical increase occurred in C1, while a similarly pronounced increase was observed in the cuboid. Finite element stress maps revealed a more spatially concentrated stress distribution in the second trimester compared with both the reference and third-trimester conditions, suggesting a transient increase in model-predicted transverse-arch stress under the second-trimester W–K input condition (Figure 3.2).

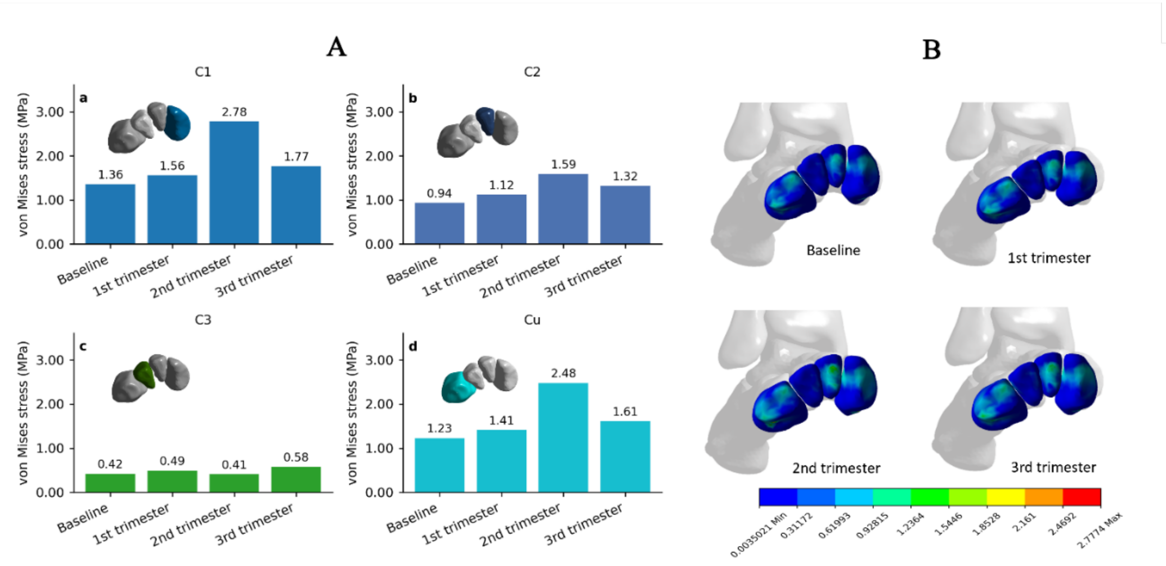


Figure 3.2. Model-predicted von Mises stress in the transverse arch region across the reference condition and the three pregnancy stages.

(A) Peak von Mises stress at the medial cuneiform (C1), intermediate cuneiform (C2), lateral cuneiform (C3), and cuboid (Cu) across reference condition and the three pregnancy stages.

(B) FE-predicted transverse-arch stress distributions across pregnancy.

3.1.3 Changes of the transverse arch during pregnancy

The transverse arch showed a non-uniform pattern of geometric change across the simulated stages of pregnancy. Although the overall profile shifted downward with gestational progression, the mediolateral and vertical components did not change in the same way. This pattern is illustrated by the trajectory of the arch apex (C2) (Figure 3.3A). The mediolateral displacement (Δx) increased first, reached a maximum of +0.19 mm in the second trimester, and then moved back toward the reference position in the third trimester. By contrast, the vertical displacement (Δy) increased progressively in the downward direction and reached −1.30 mm in the third trimester. Figure 3.3B further quantifies these changes in arch width and height. Chord length (L) increased slightly from 32.69 mm in the reference condition to 32.76 mm in the third trimester (about +0.21%), indicating a small but progressive increase in transverse span. In contrast, sagitta height (s) decreased from the reference condition to the second trimester, where it reached its lowest value (approximately 10.00 mm), and then increased slightly in the third trimester. The same trend was seen for the normalized curvature, expressed as κ/κ_ref × 100%, which decreased to 99.74% of the reference value in the first trimester, reached a minimum of 99.06% in the second trimester, and then increased to 99.51% in the third trimester. These results show that the aligned transverse-arch profile became flattest under the second-trimester input condition, whereas the greatest downward apex displacement occurred under the third-trimester input condition. The deformation vectors in Figure 3.3C show the same pattern. The largest mediolateral deviation appeared in the second trimester, while the largest inferior displacement appeared in the third trimester. Taken together, the results suggest that flattening of the transverse arch was most evident in mid-pregnancy, whereas downward displacement continued as loading increased later in gestation.

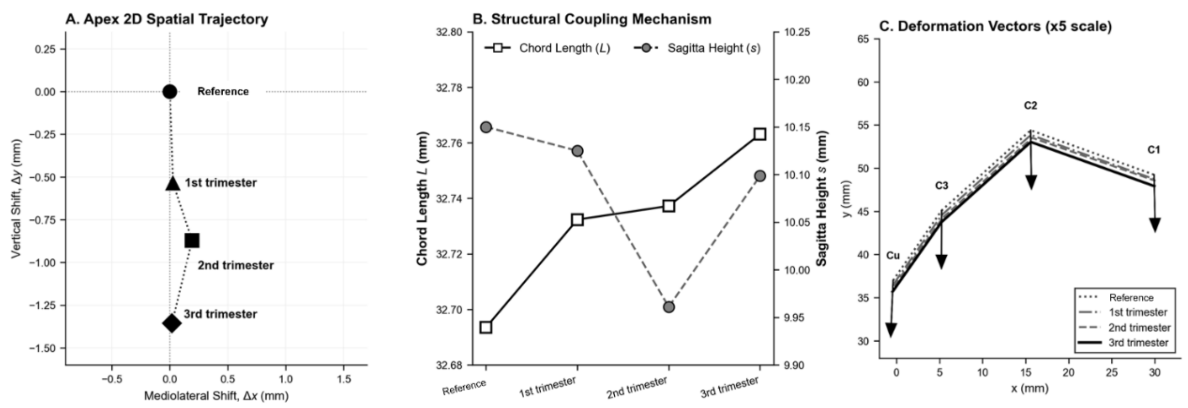


Figure 3.3. Geometric evolution of the tarsal transverse arch across pregnancy.

(A) Spatial trajectory of the arch apex relative to reference condition, illustrating stage-specific mediolateral and vertical displacement. (B) Dual-axis plot showing changes in chord length (L) and sagitta height (s) across gestation. (C) Evolution of transverse arch profiles across the four simulated stages. Deformation vectors (magnified 5×) highlight the stage-specific directions of landmark displacement relative to the reference profile.

To examine how the geometric indicators varied within the simulated W–K parameter space, standardized coefficients were obtained from the descriptive regression-based sensitivity analysis using body weight-related loading (W), passive plantar ligament–fascia stiffness-scaling parameter (K), and their interaction as predictors (Figure 3.4). For the vertical tarsal height measures (C1, C2, C3, and Cu), W showed negative coefficients ($\beta \approx -0.50$ to -0.70), whereas K showed positive coefficients ($\beta \approx +0.45$ to $+0.65$). Transverse widths showed regional differences. The C1–C2 width showed a larger coefficient for K ($\beta_K \approx +0.55$), whereas the C2–C3 and C3–Cu widths showed larger coefficients for W ($\beta_W \approx +0.50$ and $+0.95$, respectively). For the aligned transverse-arch geometry, chord length showed the largest coefficient for W ($\beta_W \approx +0.95$), while K showed a small coefficient ($|\beta_K| < 0.10$). Effective arch height showed a larger coefficient for K ($\beta_K \approx +0.55$), whereas W showed a smaller negative coefficient ($\beta_W \approx -0.25$). Relative curvature showed opposite coefficient directions for the two predictors, with a negative coefficient for W ($\beta_W \approx -0.54$) and a positive coefficient for K ($\beta_K \approx +0.47$). For apex displacement, W showed a negative coefficient in the vertical direction ($\beta_W \approx -0.40$) and a positive coefficient in the mediolateral direction ($\beta_W \approx +0.73$), whereas K showed the opposite coefficient pattern ($\beta_K \approx +0.43$ and -0.26, respectively). Interaction coefficients were generally close to zero. Overall, within the simulated design space, W showed larger coefficients for transverse widening and lateral apex shift, whereas K showed larger coefficients for effective arch height and relative curvature.

These results indicate that, under the prescribed W–K schedule, the most pronounced geometric flattening occurred in the second trimester rather than at the point of peak load. This response corresponded to the stage at which the prescribed reduction in K was greatest and should therefore be interpreted as a stiffness-linked model response rather than as an independent temporal prediction. In contrast, the greatest absolute inferior displacement of the arch apex occurred in the third trimester, indicating that stiffness-sensitive flattening and load-sensitive downward settlement did not necessarily peak at the same simulated stage.

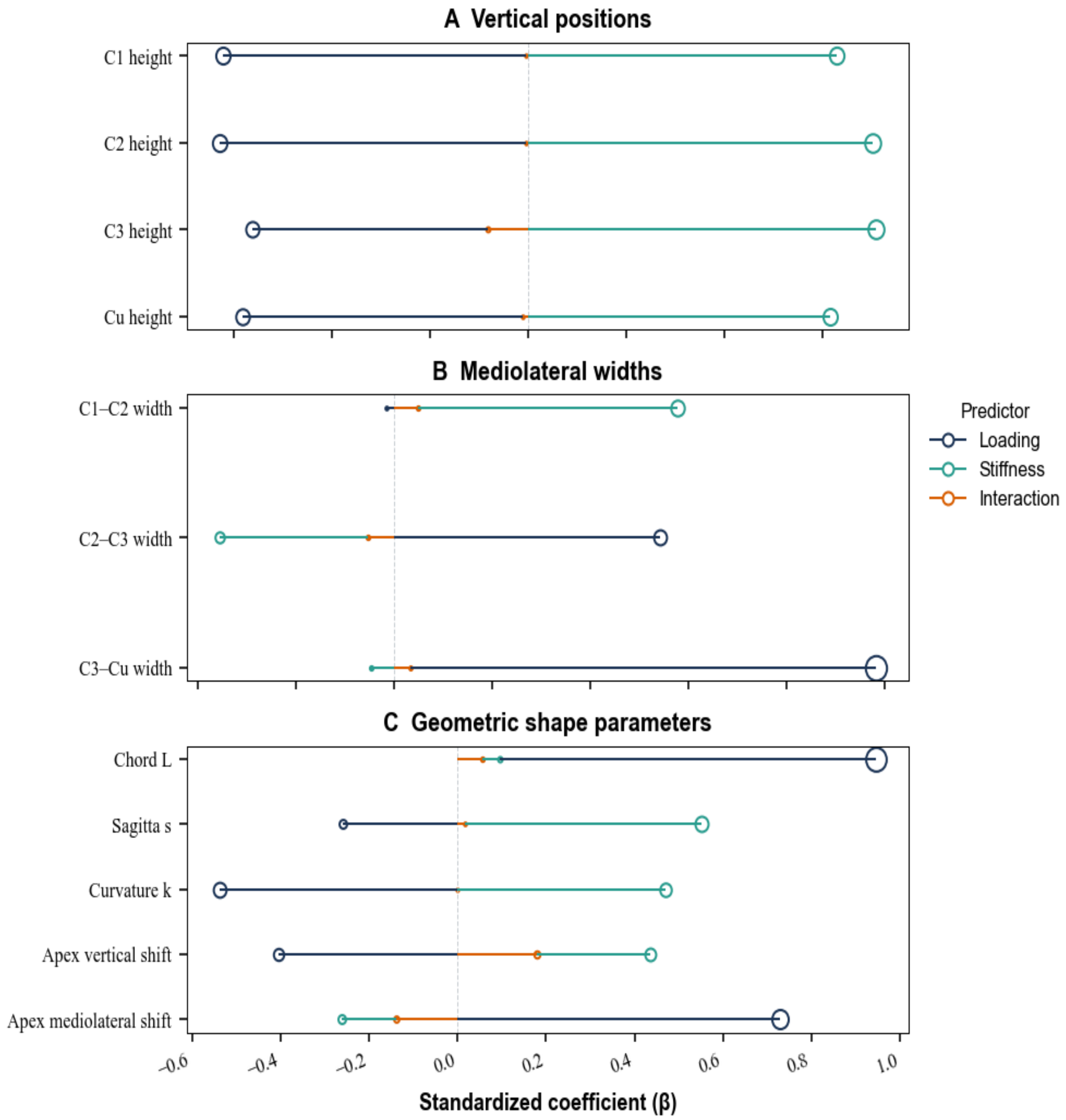


Figure 3.4. Forest plots illustrate standardized sensitivity coefficients for (A) vertical positions, (B) mediolateral widths, and (C) geometric shape parameters.

Table 3.1. Standardized sensitivity coefficients for body weight-related loading, passive plantar ligament–fascia stiffness-scaling parameter, and their interaction within the simulated W–K design space.

Outcome variable	β_W	β_K	β_interaction
Vertical heights			
C1 height	−0.70	+0.65	<0.10
C2 height	−0.60	+0.55	<0.10
C3 height	−0.55	+0.50	<0.10
Cu height	−0.50	+0.45	<0.10
Mediolateral widths			
C1–C2 width	<0.10	+0.55	<0.10
C2–C3 width	+0.50	<0.10	<0.10
C3–Cu width	+0.95	<0.10	<0.10
Shape parameters (aligned)			
Chord length (L)	+0.95	<0.10	<0.10
Sagitta (s)	−0.25	+0.55	<0.10
Curvature (κ)	−0.54	+0.47	<0.10
Apex vertical shift (Δy)	−0.40	+0.43	±0.20
Apex mediolateral shift (Δx)	+0.73	−0.26	±0.15

Note: Coefficients were obtained from a descriptive regression-based sensitivity analysis of deterministic outputs from a single subject-specific finite element model. They describe the direction and relative coefficient pattern within the simulated W–K design space and are not interpreted as population-level effect sizes or inferential statistics. Therefore, p values and corrected p values are not reported.

3.2 Gait strategies and dynamic stability analysis during pregnancy

As illustrated in Figure 3.5, the superposition of the induced accelerations, including muscle-induced, gravity-induced, and velocity-related components, reproduced the main temporal characteristics of the experimentally measured center-of-mass accelerations. This comparison was used as a qualitative consistency check of the induced-acceleration decomposition. A similar approach was used by Hamner et al., who verified that the sum of accelerations due to muscles, gravity, and velocity effects equalled the total acceleration of the body mass center [91]. In the anteroposterior, vertical, and mediolateral directions, the predicted and experimentally measured curves showed similar waveform morphology and phase characteristics, although local differences in peak magnitude were present.

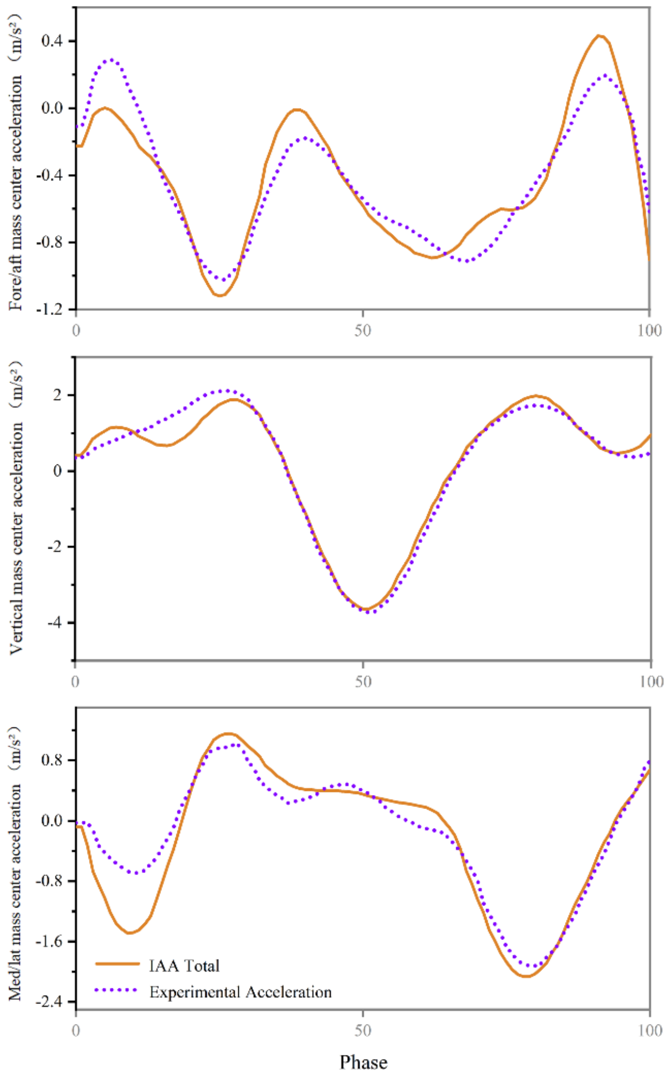


Figure 3.5. Comparison of experimentally measured and IAA-predicted center of mass accelerations in the anteroposterior, vertical, and mediolateral directions during walking.

Table 3.2 summarizes the spatiotemporal and kinematic gait parameters of pregnant and non-pregnant women. After Bonferroni correction across the seven gait-parameter comparisons, walking speed, stride length, cadence, thigh acceleration, swing acceleration, and leg impact acceleration remained significantly different between groups, whereas the single-to-double support ratio did not remain significant. Walking speed, stride length, and cadence were lower by approximately 29%, 15%, and 17%, respectively, in pregnant women. Thigh acceleration, swing acceleration, and leg impact acceleration were lower by approximately 12%, 32%, and 42%, respectively. Overall, these findings indicate a slower and less dynamically intense walking pattern in pregnant women compared with non-pregnant women.

Table 3.2. Gait parameters of non-pregnant and pregnant women.

Index	Pregnant women	CON	t	p
Walking speed (m/s)	0.83 ± 0.16*	1.17 ± 0.07	-8.71	<0.001
Stride length (m)	1.05 ± 0.07*	1.24 ± 0.04	-10.54	<0.001
Cadence (step/min)	93.41 ± 11.89*	112.47 ± 3.79	-6.83	<0.001
Thigh acceleration (g/1g = 9.8 m/s²)	1.02 ± 0.05*	1.16 ± 0.10	-5.60	<0.001
Swing acceleration (g/1g = 9.8 m/s²)	0.48 ± 0.25*	0.71 ± 0.18	-3.92	<0.001
Leg impact acceleration (g/1g = 9.8 m/s²)	1.16 ± 0.54*	2.00 ± 0.18	-6.60	<0.001
Single-to-double support ratio	3.09 ± 0.18	3.45 ± 0.50	-2.32	0.03

CON, control group, i.e., non-pregnant women. * Significant between-group difference after Bonferroni correction across the seven gait-parameter comparisons, with the corrected significance threshold set at $p < 0.0071$.

Table 3.3 summarizes the maximum and minimum stance-phase joint angles, joint range of motion (ROM), and peak sagittal-plane joint moments. After Bonferroni correction across the 15 comparisons, hip ROM remained significantly lower in pregnant women than in non-pregnant controls, whereas the between-group differences in knee and ankle ROM did not remain significant. At the hip, the minimum stance-phase angle was significantly higher in pregnant women, while the difference in the maximum angle did not remain significant after correction. At the knee, both the maximum and minimum angles were significantly lower in pregnant women, indicating a more flexed knee position during stance. At the ankle, both the maximum and minimum angles were significantly higher in pregnant women, indicating that the ankle-angle profile was shifted toward greater dorsiflexion and reduced plantarflexion.

Peak body-mass-normalized sagittal-plane joint moments were generally lower in pregnant women. Hip extension and flexion moments were lower by approximately 59% and 70%, respectively, while knee extension and flexion moments were lower by approximately 52% and 71%, respectively. The ankle dorsiflexion moment was approximately 73% lower in pregnant women. The ankle plantarflexion moment was also approximately 6% lower, although this difference did not remain significant after Bonferroni correction. Overall, these findings indicate a more flexed lower-limb posture and generally lower body-mass-normalized joint moment output during stance in pregnant women.

Table 3.3. Maximum and minimum stance-phase joint angles, joint range of motion, and peak joint moments.

	Index	Pregnant women	CON	t	p
Hip	Max stance angle (°)	46.27 ± 1.74	39.33 ± 10.41	2.94	0.01
	Min stance angle (°)	11.12 ± 2.94*	-6.61 ± 6.33	11.36	<0.001
	Stance ROM (°)	35.14 ± 3.21*	45.94 ± 9.24	-4.94	<0.001
	Peak extension moment (Nm/kg)	0.48 ± 0.08*	1.18 ± 0.60	-5.17	<0.001
	Peak flexion moment (Nm/kg)	0.28 ± 0.10*	0.94 ± 0.16	-15.64	<0.001
Knee	Max angle (°)	-19.76 ± 4.77*	-6.42 ± 3.51	-10.07	<0.001
	Min angle (°)	-69.56 ± 10.50	-59.11 ± 10.26	-3.18	≈0.002
	ROM (°)	49.77 ± 11.64	52.69 ± 12.87	-0.75	0.46
	Extension moment (Nm/kg)	0.48 ± 0.16*	1.00 ± 0.22	-8.55	<0.001
	Flexion moment (Nm/kg)	0.17 ± 0.08*	0.58 ± 0.28	-6.30	<0.001
Ankle	Max angle (°)	28.46 ± 3.63*	13.80 ± 7.14	8.19	<0.001
	Min angle (°)	-0.93 ± 10.74*	-23.52 ± 10.05	6.87	<0.001
	ROM (°)	29.39 ± 11.07	37.32 ± 10.20	-2.36	0.02
	Plantarflexion moment (Nm/kg)	1.20 ± 0.06	1.27 ± 0.08	-3.13	≈0.003
	Dorsiflexion moment (Nm/kg)	0.12 ± 0.04*	0.45 ± 0.12	-11.67	<0.001

Values are presented as mean ± SD. CON, non-pregnant control group. * Significant between-group difference after Bonferroni correction across the 15 joint-angle, ROM, and joint-moment comparisons reported in this table, with the corrected significance threshold set at $p < 0.0033$.

Table 3.4. Body-mass-normalized sagittal-plane joint work during stance in pregnant and non-pregnant women.

Variable (J/Kg)	Pregnant women	CON	p
Hip positive work	0.191 ± 0.014	0.163 ± 0.055	0.170
Hip negative work	0.016 ± 0.024	0.165 ± 0.033	<0.001*
Hip total absolute work	0.207 ± 0.019	0.328 ± 0.030	<0.001*
Knee positive work	0.079 ± 0.024	0.181 ± 0.040	<0.001*
Knee negative work	0.137 ± 0.021	0.301 ± 0.070	<0.001*
Knee total absolute work	0.216 ± 0.042	0.482 ± 0.084	<0.001*
Ankle positive work	0.160 ± 0.032	0.324 ± 0.035	<0.001*
Ankle negative work	0.198 ± 0.029	0.123 ± 0.036	<0.001*
Ankle total absolute work	0.358 ± 0.036	0.447 ± 0.043	<0.001*

Values are presented as mean ± SD. CON, non-pregnant control group. Positive work represents mechanical energy generation, whereas negative work is reported as the absolute magnitude of mechanical energy absorption. * Significant difference after Bonferroni correction for the nine joint-work comparisons, with the corrected significance threshold set at $p < 0.0056$.

Joint work analysis provided an additional energy-based description of lower-limb mechanics during stance (Table 3.4). Compared with non-pregnant controls, pregnant women showed lower hip negative work and lower hip total absolute work, whereas hip positive work did not differ significantly between groups. At the knee, positive work, negative work, and total absolute work were all lower in pregnant women. At the ankle, pregnant women showed lower positive work and lower total absolute work than controls, but higher negative work. Overall, these results indicate that late-pregnancy walking was associated with lower body-mass-normalized mechanical energy generation and exchange across most lower-limb joints, particularly reduced ankle positive work during stance.

As shown in Figure 3.6, significant difference can be observed in the continuous kinematic and kinetic data between pregnant and control groups across specific portions of the stance phase. Pregnant women exhibited significantly greater hip flexion angles during early stance (approximately 0–10% of stance) and again during terminal stance to pre-swing (approximately 55–90% of stance) ($p < 0.001$). Significant between-group differences in knee flexion angle were also observed during early stance (approximately 0–8% of stance) and late stance (approximately 58–82% of stance) ($p < 0.001$). At the ankle joint, significant differences in ankle angle were evident from mid-stance to push-off (approximately 30–95% of stance), with pregnant women showing reduced plantarflexion-related peaks compared with controls ($p < 0.001$). For the kinetic waveforms, significant group differences were observed in hip flexion moment during early stance and mid-to-late stance, in knee flexion moment during early, mid-, and late stance, and in ankle dorsiflexion moment during early stance and at the end of stance (all p < 0.001).

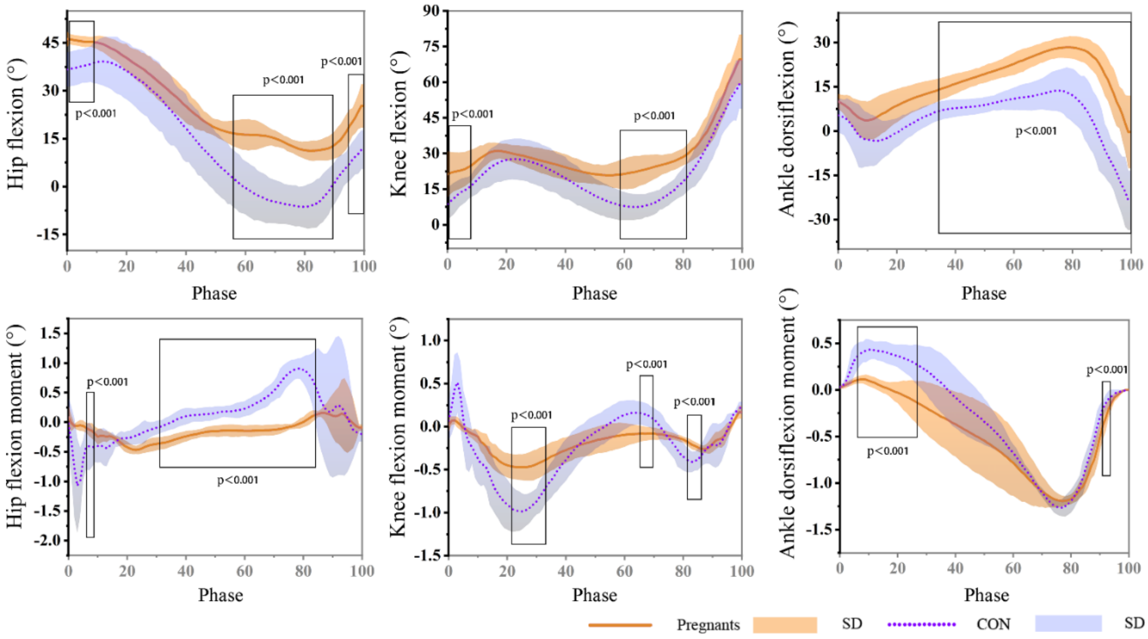


Figure 3.6. Comparison of lower-limb joint kinematics and kinetics during walking between pregnant and non-pregnant women.

The upper row shows joint angles and the lower row shows net joint moments for the hip, knee, and ankle from left to right. Orange solid lines represent pregnant women, purple dotted lines represent non-pregnant controls, and shaded areas indicate ±1 standard deviation. The stance phase was normalized from initial contact to toe-off. Rectangular boxes indicate significant between-group regions identified by SPM1d analysis.

Figure 3.7 presents the stance-averaged signed IAA-derived muscle-force-induced joint angular accelerations, reported in deg/s², at the hip, knee, and ankle. Positive values represented hip flexion, knee extension, and ankle dorsiflexion, whereas negative values represented hip extension, knee flexion, and ankle plantarflexion. The results were interpreted according to both the direction and the absolute magnitude of the induced angular acceleration. Statistical significance was determined using Bonferroni-adjusted p values across the 21 muscle-by-joint comparisons. At the hip, pregnant women showed significantly smaller magnitudes of gluteus-induced extension-related acceleration, iliopsoas-induced flexion-related acceleration, soleus-induced extension-related acceleration, and tibialis-anterior-induced flexion-related acceleration. Relative to controls, these magnitudes were lower by approximately 44.0%, 73.7%, 52.4%, and 97.8%, respectively. In contrast, the gastrocnemius-induced flexion-related acceleration was approximately 88.5% greater in pregnant women. The quadriceps- and hamstrings-induced accelerations did not remain significantly different after Bonferroni correction. At the knee, pregnant women showed a significantly smaller iliopsoas-induced flexion-related acceleration, with its magnitude reduced by approximately 60.3%. In contrast, the magnitude of the gastrocnemius-induced flexion-related acceleration was approximately 127.4% greater in pregnant women. The stance-averaged tibialis-anterior-induced acceleration changed from a flexion-related direction in controls to a small extension-related direction in pregnant women. No Bonferroni-corrected differences were observed for the gluteus, quadriceps, hamstrings, or soleus. At the ankle, the magnitude of the gastrocnemius-induced plantarflexion-related acceleration was approximately 57.0% smaller in pregnant women, while the tibialis-anterior-induced dorsiflexion-related acceleration was approximately 64.2% smaller. The iliopsoas-induced ankle acceleration changed from a dorsiflexion-related direction in controls to a small plantarflexion-related direction in pregnant women, whereas the quadriceps-induced acceleration changed from a plantarflexion-related direction in controls to a dorsiflexion-related direction in pregnant women. The soleus-induced plantarflexion-related acceleration was descriptively approximately 41.1% smaller in pregnant women, but this difference did not remain significant after Bonferroni correction.

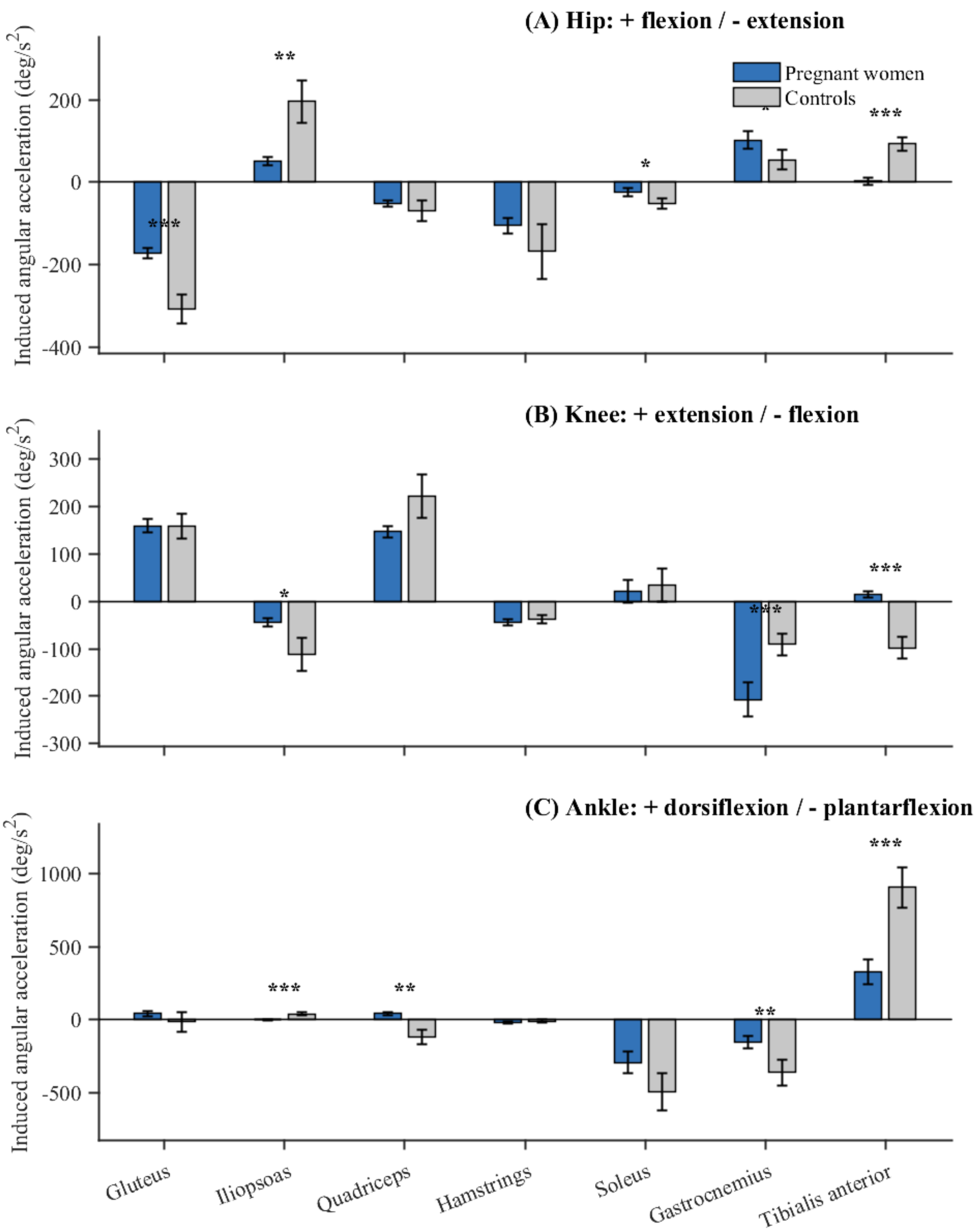


Figure 3.7. Stance-averaged signed IAA-derived muscle-force-induced joint angular accelerations in pregnant and non-pregnant women.

The signs follow the joint-coordinate conventions of the musculoskeletal model. Positive values represent hip flexion, knee extension, and ankle dorsiflexion, whereas negative values represent hip extension, knee flexion, and ankle plantarflexion. Bars represent group means, and error bars represent standard deviations. Asterisks indicate Bonferroni-adjusted p values across the 21 muscle-by-joint comparisons: * p_adj < 0.05, ** p_adj < 0.01, and *** p_adj < 0.001.

3.3 Effects of heel-toe drop on walking biomechanics

Significant between-condition differences were observed in stride length, walking speed, and single support time. Compared with the low-heel condition, stride length was approximately 11% shorter in the negative-heel condition, while the flat-shoe condition showed an intermediate value. Walking speed showed a similar pattern, with the negative-heel condition being approximately 16% slower than the low-heel condition and about 8% slower than the flat-shoe condition. Single support time was approximately 2% longer in the negative-heel condition than in the low-heel condition, with the flat-shoe condition again showing intermediate values. No significant differences were found in cadence or in the remaining spatiotemporal parameters.

Table 3.5. Spatiotemporal gait parameters under three footwear conditions during pregnancy (Mean ± SD).

Indexes (Unit)	Negative-heel shoes	Flat shoes	Low-heel shoes	p
Stride length (m)	0.99 ± 0.08c	1.05 ± 0.07	1.11 ± 0.03a	< 0.001
Walking speed (m/s)	0.76 ± 0.11bc	0.83 ± 0.16a	0.90 ± 0.08b	< 0.001
Cadence (step/min)	88.80 ± 7.80	93.60 ± 12.60	93.60 ± 8.40	0.25
Single support time (%)	75.04±0.66c	74.03±1.22	73.37±1.05a	< 0.05

Values are presented as mean ± SD. Superscript letters indicate significant post-hoc pairwise differences after Bonferroni correction: [a] compared with negative-heel shoes; [b] compared with flat shoes; [c] compared with low-heel shoes.

Figure 3.8 shows the Bonferroni-corrected SPM1D comparisons of three-dimensional ground reaction forces across the three footwear conditions during stance. Significant footwear-related differences were observed in all three GRF components, but these differences were localized to specific portions of the stance phase rather than distributed uniformly throughout stance. Compared with low-heel shoes, negative-heel shoes showed greater GRFz during 7%–9% of stance, with an average cluster difference of approximately 9.9% BW, and during 58%–70% of stance, with an average difference of approximately 4.2% BW. During 81%–85% of stance, negative-heel shoes showed lower GRFz than low-heel shoes by approximately 5.0% BW. During terminal stance and toe-off, from 92% to 100% of stance, negative-heel shoes again showed greater GRFz than low-heel shoes by approximately 5.9% BW. Negative-heel shoes also showed greater GRFz than flat shoes during 91%–100% of stance, with an average difference of approximately 5.8% BW. Flat shoes showed greater GRFz than low-heel shoes during 64%–71% of stance, with an average difference of approximately 3.0% BW.

In the anteroposterior direction (GRFy), the mean waveform showed a positive braking-related component during early stance and a negative propulsion-related component during late stance. Negative-heel shoes differed from low-heel shoes during 21%–31% and 95%–100% of stance. During 21%–31% of stance, negative-heel shoes showed a lower positive AP-GRF component than low-heel shoes by approximately 2.1% BW, indicating a smaller braking-related AP force during early-to-mid stance. During 95%–100% of stance, negative-heel shoes showed a more negative AP-GRF component than low-heel shoes by approximately 3.1% BW, indicating a greater propulsion-direction AP force at terminal toe-off. Negative-heel shoes also differed from flat shoes during 83%–88% and 95%–100% of stance. During 83%–88% of stance, negative-heel shoes showed a less negative AP-GRF component than flat shoes by approximately 2.0% BW, indicating a smaller propulsion-direction AP force during late stance. However, during 95%–100% of stance, negative-heel shoes showed a more negative AP-GRF component than flat shoes by approximately 3.0% BW, indicating a greater propulsion-direction AP force at terminal toe-off. No supra-threshold clusters were observed between flat and low-heel shoes in the AP direction after Bonferroni correction.

In the mediolateral direction (GRFx), significant differences were smaller and more localized. Compared with low-heel shoes, negative-heel shoes showed a more positive GRFx during 2%–5% and 9%–10% of stance, with average differences of approximately 1.0% BW and 1.5% BW, respectively. Compared with flat shoes, negative-heel shoes showed a more positive GRFx during 2%–5% and 8%–9% of stance, with average differences of approximately 1.1% BW and 1.3% BW, respectively. Flat shoes showed a more positive GRFx than low-heel shoes during 15%–17% and 77%–81% of stance, with average differences of approximately 1.5% BW and 1.1% BW, respectively. The corrected GRF analysis showed that footwear-related GRF differences were not distributed uniformly across the stance phase. The largest local differences were observed in the vertical direction, particularly between negative-heel and low-heel shoes. Negative-heel shoes increased vertical support by approximately 4.2%–9.9% BW during early and mid-to-late stance, but reduced GRFz by approximately 5.0% BW during 81%–85% of stance. In the AP direction, negative-heel shoes mainly altered the local AP-GRF component near toe-off by approximately 3.0%–3.1% BW. These findings indicate localized changes in the timing and distribution of stance-phase forces rather than a uniform increase or decrease in GRF throughout the entire stance phase.

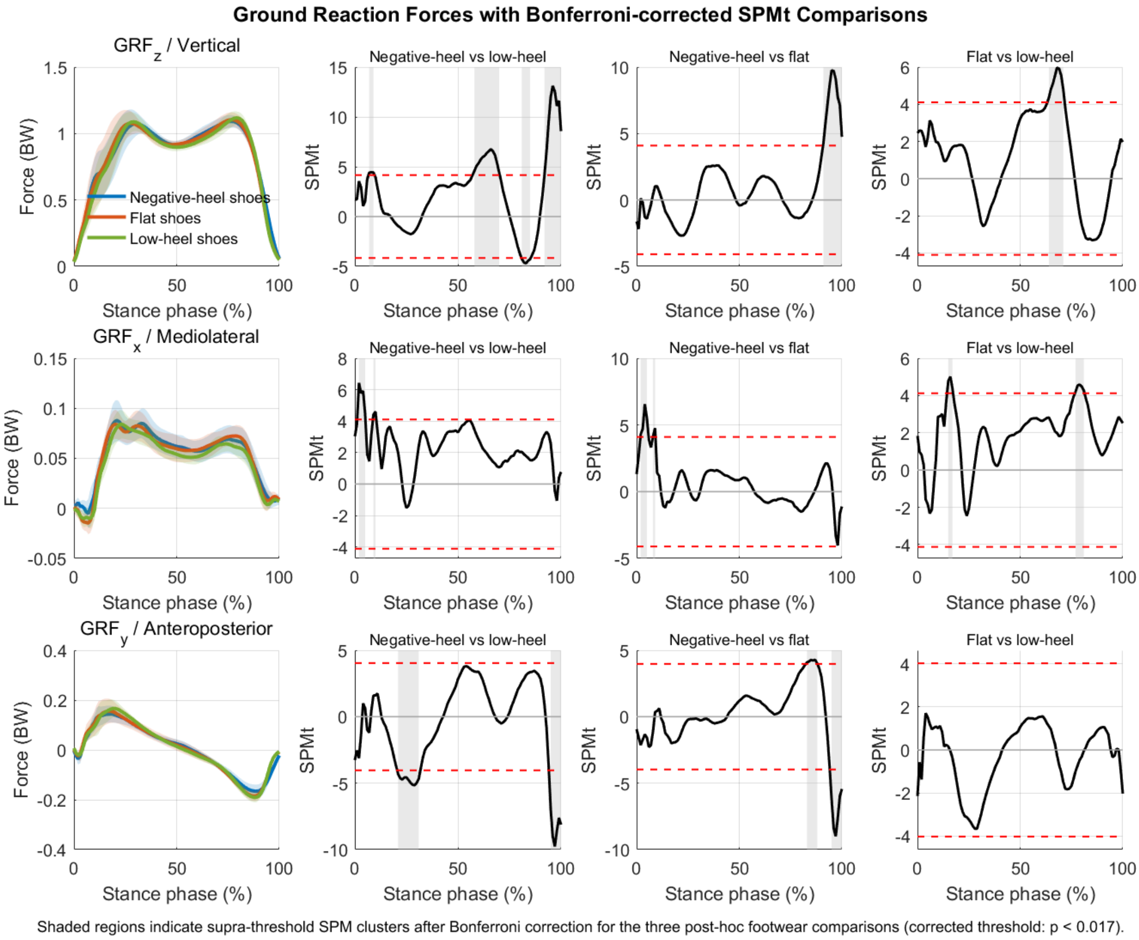


Figure 3.8. Three-dimensional ground reaction force profiles during stance phase with SPM(t) comparisons across footwear conditions.

Figure 3.9 shows the Bonferroni-corrected SPM1D comparisons of ankle sagittal-plane angle and moment across the three footwear conditions. For ankle angle, supra-threshold clusters were detected after correction in all three pairwise comparisons, but they were mainly concentrated in mid-to-late or terminal stance. Negative-heel shoes differed from low-heel shoes during 94%–99% of stance, and from flat shoes during 97% and 99%–100% of stance. Flat shoes differed from low-heel shoes during 48%, 53%–54%, and 56%–80% of stance. In contrast, no supra-threshold clusters were observed for ankle sagittal-plane moment in any pairwise comparison after Bonferroni correction.

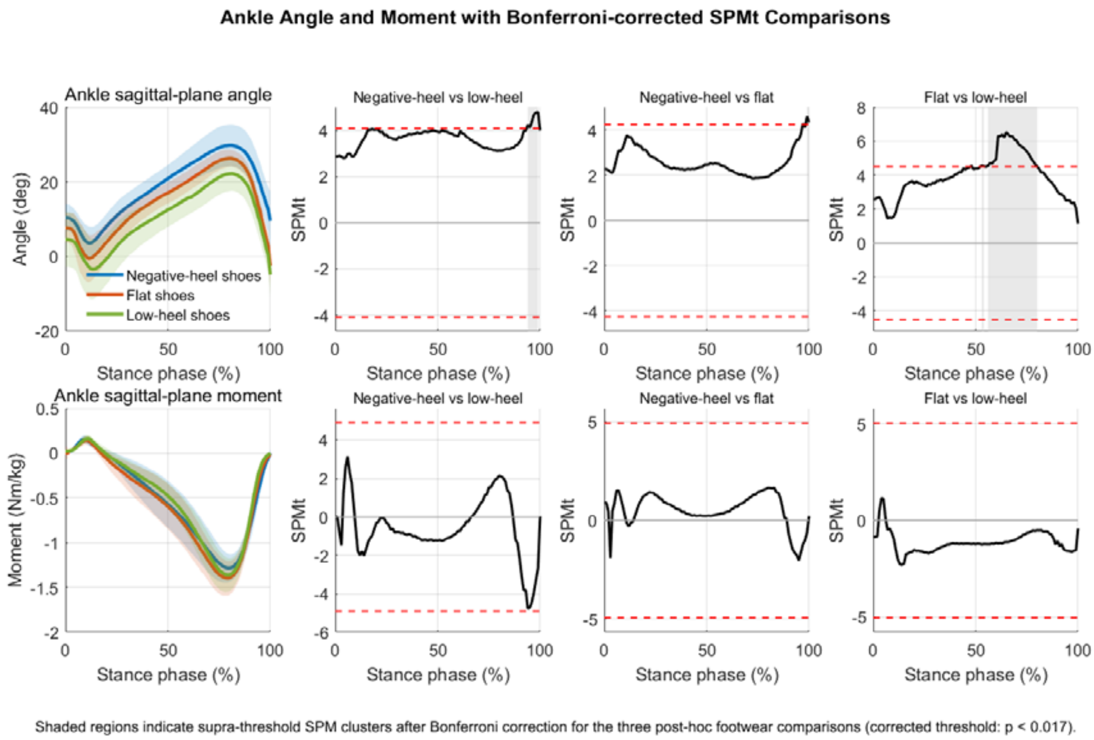


Figure 3.9. Bonferroni-corrected SPM1D comparisons of ankle sagittal-plane angle and moment across footwear conditions during stance.

Figure 3.10 shows the Bonferroni-corrected SPM1D comparisons of knee sagittal-plane angle and moment across the three footwear conditions. For knee angle, a small supra-threshold cluster was observed between negative-heel and low-heel shoes at 8% of stance. Flat shoes also differed from low-heel shoes during 74% and 76%–78% of stance. No supra-threshold cluster was detected between negative-heel and flat shoes. For knee sagittal-plane moment, no supra-threshold clusters were observed in any pairwise comparison after Bonferroni correction. These findings indicate that footwear-related knee differences were limited to short, phase-specific changes in knee angle, whereas knee moment was not significantly altered after correction.

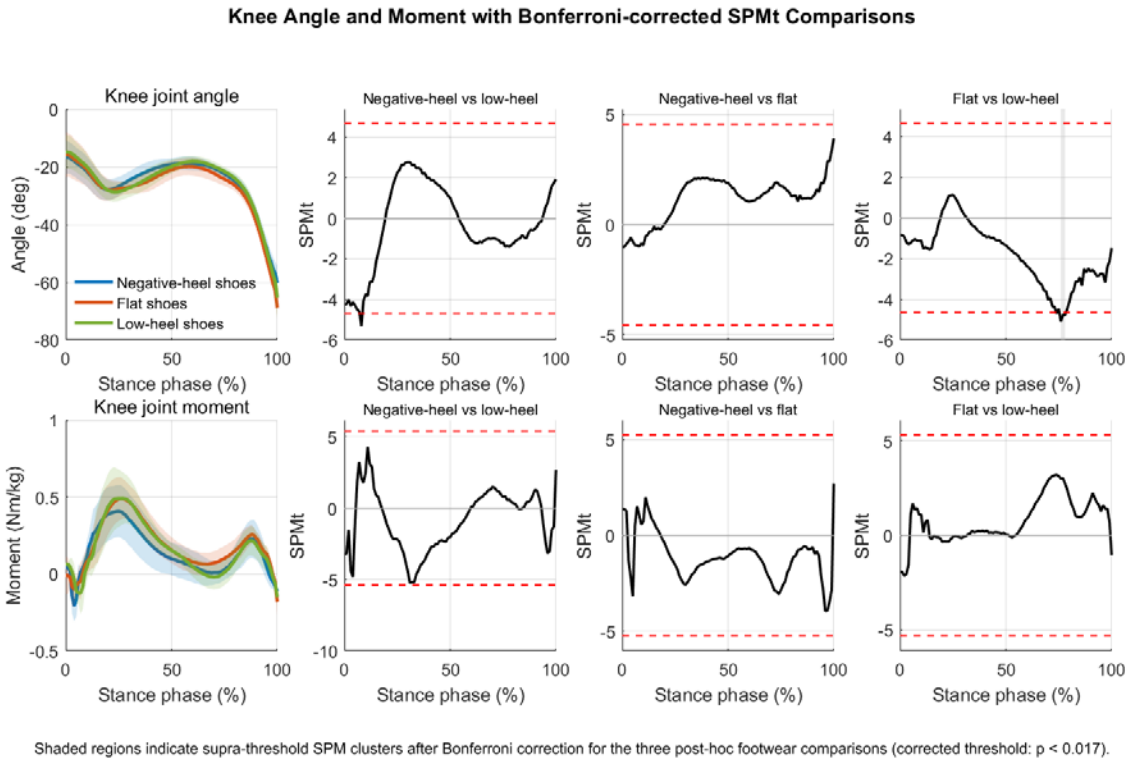
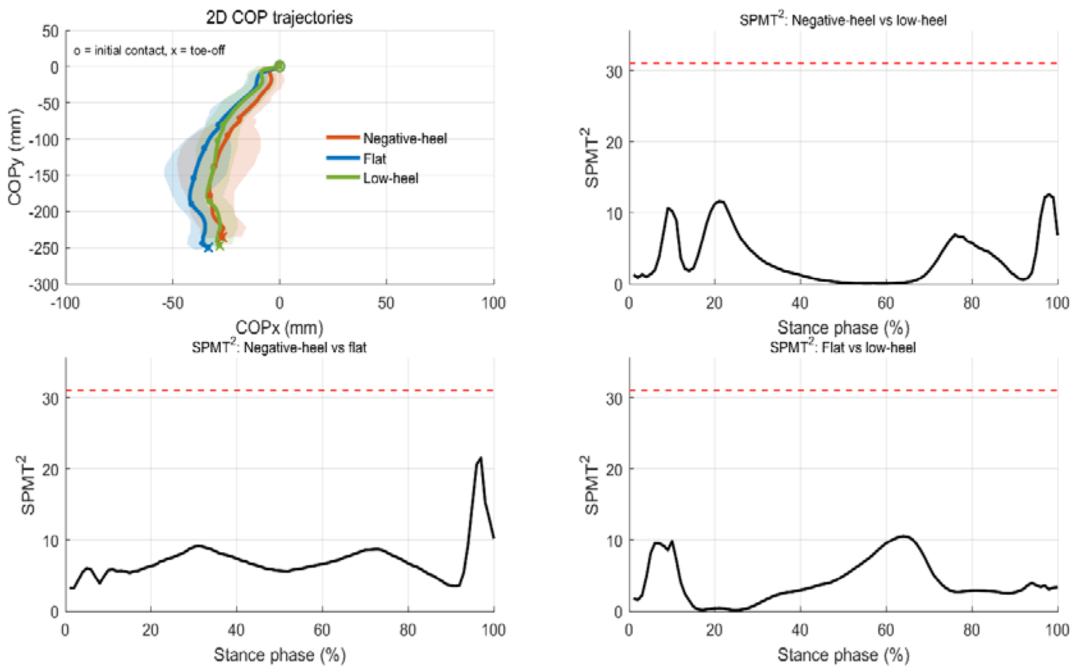


Figure 3.10. Bonferroni-corrected SPM1D comparisons of knee sagittal-plane angle and moment across footwear conditions during stance.

As shown in Figure 3.11, the mean COP trajectories showed a broadly similar forward progression pattern across the three footwear conditions. The terminal COPy displacement was approximately 237 mm in the negative-heel condition, 251 mm in the flat-shoe condition, and 249 mm in the low-heel condition. Thus, the terminal AP-related COP displacement in the negative-heel condition was descriptively about 5.5% smaller than that in the flat-shoe condition and about 4.7% smaller than that in the low-heel condition. The total COPx range was approximately 31 mm in the negative-heel condition, 40 mm in the flat-shoe condition, and 37 mm in the low-heel condition, corresponding to about 13.0%, 15.8%, and 14.9% of the terminal COPy displacement, respectively. Descriptively, the negative-heel condition therefore showed a smaller mediolateral COP excursion than the flat-shoe and low-heel conditions, with the COPx range reduced by approximately 22.8% relative to flat shoes and 17.0% relative to low-heel shoes. The peak negative COPx excursion was also smaller in the negative-heel condition than in the flat-shoe condition, by approximately 20.4%, while it was similar to the low-heel condition. The COP trajectory was further analyzed as a two-dimensional time-continuous vector field, COP(t) = [COPx(t), COPy(t)], using Bonferroni-corrected paired vector-field SPM{T^2}. No supra-threshold T^2 clusters were detected in any pairwise footwear comparison, including negative-heel versus low-heel shoes, negative-heel versus flat shoes, and flat versus low-heel shoes. Therefore, although the mean COP trajectories showed small descriptive differences, especially a smaller COPx excursion and slightly shorter terminal COPy displacement in the negative-heel condition, the overall two-dimensional COP trajectory did not differ significantly among the three footwear conditions after correction.

Two-dimensional COP trajectory and Bonferroni-corrected vector-field SPMT^2



COPx and COPy were analysed jointly as a two-dimensional trajectory. Shaded regions in the trajectory panel indicate pointwise 2D dispersion around the mean COP path. Shaded regions in SPM panels indicate supra-threshold SPMT^2 clusters after Bonferroni correction for the three post-hoc footwear comparisons (corrected threshold: $p < 0.017$).

Figure 3.11. Mean (± SD) center-of-pressure displacement in the anteroposterior and mediolateral directions during stance phase and corresponding Bonferroni-corrected vector-field SPM{T²} results for pairwise footwear comparisons.

The AP and ML COP components are shown for visualization, whereas statistical inference was performed on the two-dimensional COP vector field COP(t) = [COPx(t), COPy(t)].

As shown in Table 3.6, repeated-measures ANOVA revealed no significant effect of footwear condition on peak COP inclination angle ($F = 0.35$, $p = 0.71$) or peak forward tilt angle ($F = 1.22$, $p = 0.30$). In contrast, peak backward tilt angle differed significantly among footwear conditions ($F = 16.52$, $p < 0.01$). Post-hoc analysis showed that the negative-heel condition produced a smaller peak backward tilt angle than both the flat and low-heel conditions (both $p < 0.01$), whereas no significant difference was observed between the flat and low-heel conditions. Relative to the flat and low-heel conditions, the peak backward tilt angle in the negative-heel condition was reduced by approximately 15.8% and 11.8%, respectively.

Table 3.6. Peak centroid center of pressure inclination and tilt angles under three footwear conditions (Mean ± SD).

Indexes (Unit)	Negative-heel shoes (Mean ± SD)	Flat shoes (Mean ± SD)	Low-heel shoes (Mean ± SD)	F	p
Peak COP inclination angle (°)	3.28 ± 0.91	3.38 ± 0.70	3.14 ± 0.48	0.35	0.71
Peak COP forward tilt angle (°)	16.00 ± 1.79	15.90 ± 3.31	17.00 ± 1.61	1.22	0.30
Peak COP backward tilt angle (°)	12.82 ± 2.61[bc]	15.22 ± 2.18[a]	14.53 ± 1.72[a]	16.52	<0.01

Values are presented as mean ± SD.

a Significant difference compared with negative-heel shoes.

b Significant difference compared with flat shoes.

c Significant difference compared with low-heel shoes.

4. Discussion

This thesis examined pregnancy-related walking stability from three biomechanical perspectives: foot structural response, walking gait adaptation, and footwear-related mechanical modulation. These three components were designed to address connected but distinct aspects of the same problem. The finite element study explored how literature-informed changes in loading and passive plantar ligament–fascia stiffness could affect the mechanical response of a subject-specific foot model. The gait comparison study examined how late pregnancy was reflected in spatiotemporal parameters, lower-limb joint mechanics, joint work, and model-estimated muscle-force-induced joint angular accelerations during walking. The footwear experiment further investigated whether heel-toe drop could acutely modify stability-related gait variables in women in late pregnancy. Accordingly, the discussion is organized into three sections. Section 4.1 discusses the structural interpretation of the finite element findings and their implications for foot load transfer. Section 4.2 interprets the walking-gait results as evidence of conservative locomotor adaptation in late pregnancy. Section 4.3 discusses the acute biomechanical effects of negative-heel footwear and the extent to which these findings can be related to stability-related gait modulation. This structure follows the progression of the thesis from internal structural mechanics, to whole-body gait adaptation, and finally to an external footwear intervention.

4.1 Structural interpretation of pregnancy-related foot deformation

Within the literature-informed W–K simulation framework, the subject-specific FE model predicted changes in plantar pressure, midfoot stress, and transverse-arch geometry across the simulated pregnancy-stage conditions. These model responses were generated by prescribing trimester-specific vertical loading and passive plantar ligament–fascia stiffness-scaling parameters. Notably, experimental studies have also reported a forward redistribution of plantar load, commonly attributing the increased forefoot loading to the anterior displacement of the COM caused by the growing abdominal mass during pregnancy [33, 43]. However, within the prescribed W–K design space, the model results suggest that increased vertical loading and reduced passive plantar ligament–fascia stiffness may contribute to forefoot load redistribution even without explicitly modeling anterior COM displacement.

Previous studies provide contextual support for the plausibility of the simulated mechanisms. During pregnancy, generalized joint laxity increases markedly and is closely linked to hormonal changes [23, 24]. Existing reviews suggest that pregnancy-related endocrine changes may affect collagen-rich musculoskeletal tissues, including ligaments and fascia, but direct evidence for pregnancy-specific changes in plantar fascial stiffness remains limited [8, 22, 31]. From a mechanical perspective, the human medial longitudinal arch functions as an elastic spring, and reductions in its stiffness compromise its load-bearing and energy-storing capacity [93]. During walking, reduced arch stiffness may limit the efficiency of force transfer during push-off, which could increase the mechanical demand on the foot–ankle complex or contribute to reduced ankle push-off output. This interpretation may be relevant to the lower ankle moment observed in the pregnant group in the gait analysis, although this link was not directly tested in the static FE model. FE studies also demonstrate that decreased stiffness of the plantar soft tissues and arch-supporting structures reduces midfoot load-bearing capacity, leading to greater transfer of load towards the metatarsal heads [94]. Longitudinal pregnancy studies have further reported decreases in arch height accompanied by increases in foot width and forefoot contact area, together with elevated pressure beneath the second metatarsal head [33]. In non-pregnant populations, increased forefoot width or structural deformities similarly correlate with higher forefoot pressures, even under posture-controlled gait conditions [95]. Collectively, these findings help explain why the present study observed a relative increase in forefoot pressure despite the maintenance of constant posture, suggesting that the prescribed increase in vertical loading and reduction in passive plantar ligament–fascia stiffness may reduce arch bending stiffness and alter internal load-transfer pathways, thereby independently driving the redistribution of load towards the forefoot rather than solely reflecting an anterior shift of the COM.

Although only minor geometric changes of the tarsal transverse arch were predicted across the simulated W–K conditions, these alterations may nonetheless influence midfoot mechanics. As the transverse arch behaves as a shallow shell structure, small reductions in its curvature or height can substantially decrease transverse stiffness and modify internal load-transfer pathways [96]. Experimental and imaging studies further indicate that reductions in transverse stability diminish the midfoot's resistance to lateral splay, increasing the reliance on transverse ligamentous structures and the medial column to maintain tarsal alignment [97]. When transverse support is compromised, the forefoot becomes more prone to lateral widening, accompanied by elevated loading beneath the metatarsal heads, as demonstrated in pathological transverse-arch collapse and splay-foot conditions [98, 99]. Computational analyses also show that reduced stiffness of transverse-arch supporting structures results in increased local stresses under the metatarsal heads [94]. The present simulations are consistent with these observations and suggest that, within this subject-specific FE model, subtle reductions in transverse-arch support may alter midfoot load-sharing and increase forefoot mechanical demand.

The model-predicted transverse widening is consistent with clinical reports of increased foot length and forefoot width during pregnancy [19]. A similar pattern of transverse widening has been documented in obese individuals [100], suggesting that increased loading can be associated with mediolateral foot expansion. Consistently, within the present simulated W–K design space, W showed a stronger coefficient pattern for transverse widening than K. The descriptive sensitivity analysis indicated different response patterns within the simulated W–K design space: increased W was more closely associated with transverse widening, whereas reduced K was more closely associated with loss of effective arch height and relative curvature. These findings suggest that, within the simulated W–K design space, transverse-arch geometry responded differently to increased vertical loading and reduced passive plantar ligament–fascia stiffness. Such structural adaptations may have consequences for postural control during pregnancy. Given that pregnancy related increases in body mass and reductions in soft tissue stiffness weaken the transverse arch, the functional implications of this structural change merit consideration. The tarsal transverse arch is a key contributor to midfoot stability, providing substantial transverse stiffness that resists mediolateral deformation during weight bearing [96]. When this stiffness is reduced, the midfoot becomes more susceptible to lateral splay, which not only compromises structural support but may also alter plantar sensory input. Both mechanical stability and plantar feedback are essential components of balance regulation, and disruptions to either mechanism can affect postural control [101, 102]. These biomechanical relationships suggest that even subtle reductions in transverse-arch integrity could influence balance robustness during pregnancy, linking midfoot structural changes with broader functional demands on the foot–ankle complex.

From a mechanical standpoint, these structural changes are relevant to weight-bearing behavior of the foot because the transverse arch contributes to midfoot stiffness [96]. A reduction in transverse-arch height or curvature may decrease the resistance of the midfoot to deformation and may increase the tendency for compensatory frontal-plane foot motion, including greater inversion–eversion adjustment during weight bearing. Such changes at the foot level could influence the orientation of the shank and the alignment of the lower limb, and may therefore initiate posture adjustments beyond the foot itself. In this sense, transverse-arch deformation is not only a local structural change, but may also represent a distal mechanical factor capable of influencing whole-body postural organization during pregnancy.

The model was based on a single-subject geometry. Changes in tissue stiffness were introduced parametrically and were not measured in vivo. The results should therefore be interpreted as subject-specific mechanical findings. They do not establish population-level relationships, and they do not permit direct inference from structural deformation to balance impairment or gait instability.

4.2 Biomechanical interpretation of walking gait during pregnancy

This study compared walking gait differences between women in late pregnancy and non-pregnant controls and further applied induced acceleration analysis to estimate IAA-derived muscle-force-induced joint angular accelerations at the hip, knee, and ankle during stance. In this thesis, muscle-induced joint angular acceleration refers to the joint angular acceleration induced by selected muscle-group forces in the musculoskeletal model. These values were interpreted as indicators of the kinetic influence of muscle-group forces on joint motion, rather than as direct measurements of muscle activation, muscle strength, or neural control. The results showed that women in late pregnancy walked with lower speed, shorter stride length, lower cadence, reduced segmental accelerations, and lower sagittal-plane joint moments than non-pregnant controls. Together, these measured gait and kinetic findings suggest that late-pregnancy walking was characterized by a lower-output and more conservative gait pattern. IAA further showed that the gastrocnemius produced a smaller plantarflexion-related acceleration at the ankle but greater flexion-related accelerations at the knee and hip, while the tibialis anterior produced a smaller dorsiflexion-related acceleration at the ankle.

In terms of spatiotemporal parameters, and consistent with established literature [103, 104], walking speed, stride length, and cadence were significantly lower in the pregnant group than in the control group. This pattern, characterized by reduced velocity and shortened steps, is consistent with a more conservative walking pattern that may reduce balance-related challenge during gait [46]. Furthermore, significant reductions in thigh acceleration, swing acceleration, and leg impact acceleration were observed among the pregnant participants. These findings indicate a lower-impact walking pattern, achieved through the deceleration of limb swing and the modulation of landing impact forces, thereby reducing excessive oscillation of the musculoskeletal system during walking. The single-to-double support ratio was numerically lower in the pregnant group, but this difference did not remain significant after Bonferroni correction. Therefore, this variable should not be interpreted as a robust between-group difference in the present dataset. Nevertheless, the overall pattern of slower walking speed, shorter stride length, lower cadence, and reduced segmental accelerations is consistent with a more conservative gait strategy in late pregnancy. Prior research indicates that these adaptive modifications facilitate increased gait stability and reduce the forces generated throughout the gait cycle [105, 106]. Such a gait pattern is posited as important for the maintenance of stability [46, 105, 107] and may also be associated with reduced mechanical demand during locomotion [105, 108].

From a kinetic perspective, pregnant women showed lower body-mass-normalized sagittal-plane joint moment output during walking. This pattern should be interpreted cautiously because the values were normalized to body mass and the pregnant group also walked more slowly with shorter stride length. Therefore, the results indicate lower normalized joint moment output under the tested walking condition, rather than proving that the absolute muscular demand of walking was lower in pregnancy. Although the reduction in ankle plantarflexion moment was smaller than that observed at the hip and knee, the lower ankle positive work and smaller gastrocnemius-induced plantarflexion-related angular acceleration suggest a reduced distal push-off-related mechanical effect during stance. The soleus-induced plantarflexion-related acceleration was also smaller in magnitude, although this difference did not remain significant after Bonferroni correction.

The additional joint work analysis provides an energy-based interpretation of the joint-level kinetic findings. Compared with peak joint moments alone, joint work describes the cumulative mechanical energy generated or absorbed by a joint during stance. The most relevant finding was the lower ankle positive work in pregnant women, which indicates reduced ankle energy generation during stance and supports the interpretation of reduced distal push-off contribution in late pregnancy. This is consistent with the slower walking speed, shorter stride length, lower ankle moment output, and smaller gastrocnemius-induced plantarflexion-related angular acceleration observed in the pregnant group. However, the joint work results do not support a simple interpretation that ankle function was globally reduced. Pregnant women showed higher ankle negative work, indicating greater mechanical energy absorption or controlled energy dissipation at the ankle during stance. This suggests that the ankle may shift from a predominantly energy-generating role toward a relatively greater energy-absorbing role during late-pregnancy walking. At the knee, the lower positive, negative, and total work indicate reduced mechanical energy exchange, which is consistent with the lower movement intensity and reduced joint moment output observed in the pregnant group. At the hip, positive work did not differ significantly between groups, whereas negative and total work were lower, suggesting that the joint work results do not provide evidence of increased hip energy generation as a compensatory mechanism. These findings should be interpreted as body-mass-normalized joint work rather than absolute joint work, because the joint moments used for the calculation were normalized to body mass. Therefore, lower J/kg values in pregnant women may partly reflect the greater maternal body mass used for normalization, in addition to slower walking speed, shorter stride length, and altered joint motion. Within this limitation, the joint work results suggest that late-pregnancy walking is characterized by joint-specific changes in stance-phase mechanical energy exchange, particularly reduced distal energy generation at the ankle and reduced mechanical energy exchange at the knee.

Induced acceleration analysis provided additional information on how the estimated forces of selected muscle groups mechanically induced angular acceleration at the hip, knee, and ankle. Because the pregnancy-adjusted model was driven by experimentally measured kinematics and ground reaction forces, these results quantify the mechanical effects of the estimated muscle-group forces under the observed walking conditions. At the ankle, pregnant women showed a smaller gastrocnemius-induced plantarflexion-related acceleration and a smaller tibialis-anterior-induced dorsiflexion-related acceleration. The soleus-induced plantarflexion-related acceleration was also smaller, although the difference did not remain significant after Bonferroni correction. Together with the lower ankle positive work, the smaller gastrocnemius-induced plantarflexion acceleration supports reduced propulsion-related mechanical output at the ankle. However, ankle function was not globally reduced, because ankle negative work was greater in pregnant women, indicating greater mechanical energy absorption during stance. The gastrocnemius showed a different pattern at the knee and hip. Although its plantarflexion-related effect at the ankle was smaller, its flexion-related effects at the knee and hip were greater in pregnant women. The greater knee-flexion-related effect is consistent with the more flexed knee posture observed during stance. Because the gastrocnemius does not cross the hip, its greater hip-flexion-related effect demonstrates that a muscle force can influence joints that it does not directly cross through multibody dynamic coupling and the foot–ground constraint. Therefore, the smaller ankle effect should not be interpreted as a general reduction in gastrocnemius function. Instead, the estimated gastrocnemius force produced less plantarflexion acceleration at the ankle but greater flexion acceleration at the knee and hip. At the hip, the gluteus-induced extension-related acceleration and iliopsoas-induced flexion-related acceleration were both smaller in pregnant women. Together with the lower hip moments and lower hip negative and total work, these findings indicate reduced mechanical effects in both principal sagittal directions at the hip. Because hip positive work did not increase, the results do not support a simple proximal compensation in which reduced ankle propulsion was replaced by greater hip energy generation. The direction changes observed for the tibialis-anterior-induced knee acceleration and the iliopsoas- and quadriceps-induced ankle accelerations should not be interpreted as reversals of the anatomical functions of these muscles. They indicate that changes in joint posture, segmental inertial properties, external loading, and dynamic coupling altered the net joint acceleration produced by the estimated muscle forces. The principal IAA finding was that pregnant women showed smaller gastrocnemius- and tibialis-anterior-induced acceleration effects at the ankle, while the gastrocnemius simultaneously produced greater flexion-related effects at the knee and hip. The results therefore indicate reduced distal propulsion-related mechanics together with altered transmission of gastrocnemius force across the lower-limb kinetic chain.

Several limitations define the scope of this interpretation. First, the comparison was cross-sectional rather than longitudinal, so the observed differences cannot be interpreted as within-subject progression across pregnancy. Second, IAA provided model-derived estimates of muscle-induced joint angular acceleration rather than direct physiological measurements such as EMG. Third, the analysis was confined to the sagittal plane and therefore does not capture the full three-dimensional complexity of gait adaptation during pregnancy. Within these limits, the present findings indicate that late pregnancy was associated with lower movement intensity, reduced distal propulsive function during late stance, and altered relative weighting of synergistic muscles across the hip, knee, and ankle.

4.3 Biomechanical effects of negative-heel footwear in late pregnancy

The footwear experiment showed that heel-toe drop can acutely modify walking biomechanics in late pregnancy. Compared with flat and low-heel shoes, the negative-heel condition was associated with lower self-selected walking speed, shorter stride length, longer single-support time, altered timing of stance-phase ground reaction forces, phase-specific changes in knee and ankle angles, and a smaller peak backward COM–COP tilt angle. These results indicate that negative-heel shoes did not merely modify local foot posture, but altered the overall organization of late-pregnancy walking under self-selected speed conditions.

Together with the findings from the second part of this thesis, late-pregnancy walking was already characterized by slower walking speed, shorter stride length, lower segmental accelerations, lower sagittal-plane joint moments, and lower ankle positive work. These changes indicate a lower-output and more conservative gait pattern, in which forward progression intensity and lower-limb mechanical demand are reduced. In late pregnancy, such a pattern may represent an adaptive response to increased body mass, altered mass distribution, and greater postural-control demands. Previous pregnancy gait studies have also interpreted slower walking speed and shorter stride length as part of conservative gait organization during pregnancy (Foti et al., 2000; Branco et al., 2013; Błaszczyk et al., 2016). The negative-heel condition produced changes in the same general direction, including lower walking speed, shorter stride length, and longer single-support time. However, this similarity should not be interpreted as evidence that the same mechanism was responsible. Pregnancy-related conservative gait is mainly driven by maternal anatomical and mechanical changes, whereas the negative-heel response is more likely related to footwear geometry. By placing the heel lower than the forefoot, negative-heel shoes alter the heel-to-toe relationship and may place the ankle-foot complex in a relatively more dorsiflexed orientation during stance. Although the footwear-related ankle-angle difference was not statistically significant after correction, this descriptive tendency is mechanically consistent with the expected effect of negative heel-to-toe geometry. A relatively more dorsiflexed stance posture may modify initial foot-ground contact, foot rollover, and the timing of body progression over the supporting foot. Previous footwear biomechanics studies have shown that shoe structure and heel-to-toe configuration can influence foot rollover, sagittal-plane alignment, and stance-phase loading (Myers et al., 2006; Janisse and Janisse, 2008; Chapman et al., 2013; Shih et al., 2013). Therefore, the combination of lower walking speed, shorter stride length, longer single-support time, and the tendency toward greater stance-phase dorsiflexion under the negative-heel condition can be interpreted as an acute footwear-induced reorganization superimposed on the already conservative gait pattern of late pregnancy. In other words, negative-heel shoes did not appear to increase walking output or propulsive capacity. Instead, they may have externally reinforced a low-output walking strategy by reducing forward progression intensity and altering the ankle-foot rollover condition and temporal distribution of stance-phase support. This provides a biomechanical context for the accompanying GRF and COM–COP findings: the footwear condition likely changed how the body moved over the base of support during stance, rather than merely changing local foot posture.

The most relevant GRF changes were observed in the vertical and anteroposterior components during the late part of the normalized stance phase. In the present analysis, the stance phase was normalized from initial contact to toe-off. Negative-heel shoes reduced vertical GRF around 81%–85% of stance but increased the propulsion-direction AP-GRF near 95%–100% of stance. This combination suggests that the limb showed reduced vertical support demand at the end of terminal stance, while the forward-directed force was concentrated during the final pre-swing/toe-off portion. Biomechanically, this pattern may be related to the negative heel-to-toe geometry. By placing the heel lower than the forefoot, negative-heel shoes may alter foot rollover and the timing of body progression over the supporting foot. This geometry may also maintain the ankle in a relatively more dorsiflexed position during the middle-to-late stance phase, thereby increasing the stretch or passive tension of the gastrocnemius–soleus–Achilles tendon complex before push-off. From a muscle–tendon mechanics perspective, such pre-stretch could contribute to greater elastic energy storage and subsequent release during the final toe-off period, or may require altered plantar-flexor force production to complete forward progression. The reduced vertical GRF at 81%–85% of stance may indicate that the stance limb began to unload or required less vertical support before toe-off. In contrast, the increased propulsion-direction AP-GRF at 95%–100% of stance suggests that the remaining forward progression was expressed mainly at the final toe-off-related period. Together with the slower walking speed and shorter stride length, this pattern suggests a more cautious gait strategy in which overall forward progression is reduced, but the final propulsive force is more temporally concentrated near toe-off. However, because walking speed differed among footwear conditions, these GRF differences cannot be attributed solely to footwear geometry. Moreover, because muscle activation and tendon strain were not measured directly in the present study, the proposed gastrocnemius–soleus–Achilles tendon mechanism should be regarded as a plausible biomechanical explanation rather than direct evidence of neuromuscular or tendon-level adaptation.

The COM–COP result was the most direct stability-related finding in the footwear experiment. The peak backward COM–COP tilt angle was significantly lower in the negative-heel condition than in both the flat and low-heel conditions, whereas the overall peak COM–COP inclination angle and peak forward COM–COP tilt angle did not differ significantly among the three footwear conditions. The COM–COP relationship reflects the position of the body mass relative to the base of support and is therefore relevant to whole-body balance control during standing and walking (Winter, 1995; Pai and Patton, 1997; Hof et al., 2005). In the present study, the smaller backward COM–COP tilt angle indicates that negative-heel shoes reduced the posterior inclination of the COM relative to the COP during stance. This selective change suggests that negative-heel shoes did not produce a generalized alteration in all COM–COP measures, but mainly influenced backward sagittal-plane alignment. Together with the slower walking speed, shorter stride length, and altered AP-GRF timing, this finding suggests that negative-heel shoes may encourage a more conservative sagittal-plane body–support relationship during late-pregnancy walking.

Overall, negative-heel shoes acutely modified self-selected gait pattern, stance-phase support and braking–propulsion timing, lower-limb joint posture, and backward COM–COP alignment in late pregnancy. Their effect was not characterized by increased Bonferroni-corrected knee or ankle moment output, but rather by stability-related gait modulation through speed adjustment, postural reorganization, and redistribution of external force timing. Therefore, heel-toe drop can be interpreted as a relevant external mechanical factor capable of acutely modifying late-pregnancy walking biomechanics. However, because the present experiment measured only acute biomechanical responses during walking and did not record fall events or long-term adaptation, these findings should not be interpreted as direct evidence that negative-heel shoes provide a clinical fall-prevention effect.

5. Conclusions and future work

5.1 Structural adaptation of the foot arch during pregnancy

The first hypothesis proposed that increased body weight-related loading and reduced passive plantar ligament–fascia stiffness would be associated with distinct transverse-arch deformation patterns within a subject-specific finite element framework. Within the prescribed W–K simulation design, the model indicated a mechanical separation between load-sensitive transverse widening and stiffness-sensitive arch flattening. Increased vertical loading was more closely associated with transverse arch widening and lateral apex displacement, whereas reduced passive plantar ligament–fascia stiffness was more closely associated with loss of effective arch height and relative curvature. The accompanying increase in midfoot stress suggests that these geometric changes may alter internal load transfer within the modeled foot. Thus, the main value of this analysis is that it separates two mechanically concurrent pregnancy-related factors that are difficult to isolate in vivo and shows that they may influence transverse-arch geometry through different response patterns. These findings should be interpreted as subject-specific model-based evidence rather than population-level conclusions. Future studies should extend this framework to larger subject-specific samples and longitudinal datasets with directly measured pregnancy-stage loading and plantar soft-tissue properties.

5.2 Biomechanical interpretation of conservative walking gait adaptation in late pregnancy

The second hypothesis proposed that the conservative gait pattern observed in late pregnancy was associated with altered muscle-force-induced joint acceleration patterns, rather than representing only a passive consequence of structural change. This hypothesis was supported by the combined gait and modeling results. Compared with non-pregnant controls, women in late pregnancy walked with lower speed, shorter stride length, lower cadence, reduced segmental accelerations, and lower sagittal-plane joint moments. These findings indicate that late-pregnancy gait was characterized by reduced movement intensity and lower joint moment output.

The induced acceleration analysis showed that pregnant women had a smaller gastrocnemius-induced plantarflexion-related acceleration and a smaller tibialis-anterior-induced dorsiflexion-related acceleration at the ankle. The soleus-induced plantarflexion-related acceleration was also smaller in magnitude but did not remain significant after Bonferroni correction. In contrast, the gastrocnemius produced greater flexion-related accelerations at the knee and hip. These findings indicate that late-pregnancy walking involved reduced propulsion-related acceleration at the ankle, while the mechanical effect of the gastrocnemius was directed more strongly toward flexion-related acceleration at the knee and hip. These results were derived from experimentally measured gait data using a pregnancy-adjusted musculoskeletal model. They therefore provide quantitative evidence that the estimated gastrocnemius force produced less plantarflexion-related acceleration at the ankle but greater flexion-related acceleration at the knee and hip during late-pregnancy walking. The IAA results describe the mechanical effects of model-estimated muscle forces and should not be interpreted as direct measurements of muscle activation or neural control. Future work should combine time-resolved IAA with joint angular velocity, muscle–tendon and muscle-fiber velocity, muscle and tendon power, and electromyography. This would clarify whether the observed differences arise from altered muscle-force magnitude, contraction timing, joint configuration, dynamic coupling, or their interaction.

5.3 Footwear intervention and modulation of gait stability

The third hypothesis proposed that structurally designed footwear, specifically NHS, would modify gait mechanics in late pregnancy through measurable changes in lower-limb loading and alignment. The experimental results partly supported this interpretation, showing that NHS were associated with a smaller posterior COM-COP inclination angle, greater knee flexion at initial contact, and Negative-heel shoes produced a phase-specific redistribution of the anteroposterior ground reaction force, including a more concentrated propulsion-direction force near terminal toe-off. These findings suggest that NHS were associated with altered sagittal-plane gait behaviour in late pregnancy. However, walking speed also differed among footwear conditions, with participants walking more slowly under the NHS condition. Because walking speed can influence COM-COP inclination, ground reaction forces, and lower-limb joint biomechanics, the observed differences should be interpreted as the combined effect of footwear geometry and self-selected speed adaptation rather than the isolated effect of NHS. Within this context, the main adjustment appeared to involve reduced distal propulsive demand together with greater knee involvement during initial contact, rather than enhanced propulsion through greater ankle output. Therefore, the present findings provide preliminary biomechanical evidence that negative-heel footwear can modify gait mechanics in late pregnancy, but they do not establish whether these changes translate into improved gait stability or reduced fall risk. Before the practical value of negative-heel footwear can be fully established, its longer-term effects, speed-controlled responses, user comfort, adherence, and real-world safety should be evaluated in larger cohorts of pregnant women.

Thesis points

1st Thesis point:

Using a subject-specific three-dimensional finite element model, this study quantitatively evaluated the mechanical contributions of increased body weight and reduced soft-tissue stiffness to pregnancy-related foot deformation. The descriptive sensitivity analysis indicated that increased body weight was more strongly associated with mediolateral widening of the transverse arch, whereas reduced tissue stiffness was more strongly associated with vertical lowering of the modeled arch structure. Within the analyzed subject-specific model, these geometric changes were accompanied by increased internal stress in the midfoot, with the largest numerical increase occurring in the medial cuneiform and a similarly pronounced increase occurring in the cuboid under the simulated pregnancy conditions. These findings indicate that, in addition to whole-body postural changes, the combined effects of loading and tissue compliance may contribute to altered midfoot mechanics during pregnancy. This result should be interpreted as a subject-specific mechanistic finding derived from finite element modeling.

Related articles to the 1st thesis point:

1. **X Li**, Z Lu, András, K., Gusztáv, F., & Gu, Y. (2025). Biomechanical impact of weight gain and ligament relaxation on arch deformation during pregnancy: a Finite Element Analysis. Footwear Science, 17(sup1), S146-S147.

2nd Thesis point:

Induced acceleration analysis demonstrated altered IAA-derived muscle-force-induced joint angular acceleration patterns across the lower-limb joints during walking in late pregnancy. Across the hip, knee, and ankle joints, the magnitudes of muscle-induced acceleration were generally lower in pregnant women than in non-pregnant controls, indicating an overall lower-output gait pattern. At the ankle, gastrocnemius-induced plantarflexion-related acceleration was significantly smaller in pregnant women, while soleus-induced acceleration was descriptively smaller but did not remain significant after correction. At the knee, the contribution pattern became less clearly quadriceps-dominant in pregnant women, indicating a redistribution of IAA-derived joint angular acceleration estimates across the lower limb. Taken together, these model-based findings suggest that pregnancy-related walking is characterized by reduced distal contribution and a redistribution of IAA-derived angular acceleration patterns, with the clearest redistribution observed at the ankle. When interpreted together with the reduced walking speed, shorter stride length, and lower joint moments observed in this study, this pattern is biomechanically consistent with a more conservative and stability-oriented gait strategy. Because these conclusions are derived from induced acceleration analysis and musculoskeletal modeling, they should be interpreted as biomechanical inferences rather than as direct evidence of neuromuscular regulation.

Related articles to the 2nd thesis point:

1. **X Li**, Z Lu, Y Song, M Liang, Y Yuan, G Fekete, A Kovács, D Sun, Y Gu. (2024). Pregnancy-induced gait alterations: meta-regression evidence of spatiotemporal adjustments. Frontiers in bioengineering and biotechnology, 12, 1506002. IF: 4.8, Q1.
2. **X Li**, Z Lu, X Cen, Y Zhou, R Xuan, D Sun, Y Gu. (2023). Effect of pregnancy on female gait characteristics: A pilot study based on portable gait analyzer and induced acceleration analysis. Frontiers in Physiology, 14, 1034132. IF: 3.4, Q1.

3rd Thesis point:

This study demonstrated that NHS were associated with measurable changes in gait mechanics during late pregnancy. Compared with flat and low-heel shoes, the negative-heel condition reduced the peak posterior COM-COP inclination angle to 12.82°, which was lower than the values observed in the flat-shoe condition (15.22°) and the low-heel condition (14.53°). In addition, NHS were associated with greater knee flexion at initial contact and with altered ground reaction force patterns during stance, particularly in the anteroposterior and vertical directions. The reduced backward tilt angle, together with the modified propulsion-related force profile, indicates that NHS changed sagittal-plane gait mechanics in a manner consistent with a more conservative walking pattern during late pregnancy. From a biomechanical perspective, these findings suggest that the use of NHS may reduce posteriorly directed COM-COP excursion demands and redistribute mechanical loading during walking. However, because the present results reflect short-term biomechanical responses and do not include direct fall outcomes, the implications for fall prevention should be interpreted indirectly. Overall, the findings suggest that NHS may be a potentially useful footwear design for modifying gait stability-related biomechanical indicators in late pregnancy.

Related articles to the 3rd thesis point:

1. X Li, Z Lu, D Sun, R Xuan, Z Zheng, Y Gu. (2022). The influence of a shoe’s heel-toe drop on gait parameters during the third trimester of pregnancy. Bioengineering, 9(6), 241. IF: 3.7, Q2.

List of publications

Peer-reviewed articles related to this thesis:

1. **X Li**, Z Lu, Y Song, M Liang, Y Yuan, G Fekete, A Kovács, D Sun, Y Gu. (2024). Pregnancy-induced gait alterations: meta-regression evidence of spatiotemporal adjustments. Frontiers in bioengineering and biotechnology, 12, 1506002. **IF: 4.8, Q1**.
2. **X Li**, Z Lu, X Cen, Y Zhou, R Xuan, D Sun, Y Gu. (2023). Effect of pregnancy on female gait characteristics: A pilot study based on portable gait analyzer and induced acceleration analysis. Frontiers in Physiology, 14, 1034132. **IF: 3.4, Q1**.
3. **X Li**, Z Lu, D Sun, R Xuan, Z Zheng, Y Gu. (2022). The influence of a shoe's heel-toe drop on gait parameters during the third trimester of pregnancy. Bioengineering, 9(6), 241. **IF: 3.7, Q2**.
4. Z Lu, **X Li**, D Sun, Y Song, G Fekete, A Kovács, Z Gao, J Zheng, L Xiang, Y Gu. (2025). Computationally tuned dual-layer lattice pads adapted to gait-induced pressure distribution. npj Advanced Manufacturing, 2(1), 43. **Nature Portfolio journal**.
5. Jiang X, **Li X**, Xu Y, et al. Can PAPE-induced increases in jump height be explained by jumping kinematics? Molecular & Cellular Biomechanics, 2023, 20(2): 67. **IF: 1.0, Q4**
6. Z Lu, **X Li**, D Sun, Y Song, G Fekete, A Kovács, K András, Y Gu. (2025). Parametric cushioning lattice insole based on finite element method and machine learning: A preliminary computational analysis. Journal of Biomechanics, 184, 112674. **IF: 2.4, Q1**.
7. Z Lu, **X Li**, D Sun, Y Song, G Fekete, A Kovács, Y Gu. (2025). Will this be the next step? A systematic review of 3D printing in footwear biomechanics. Footwear Science, 17(2), 127-142. **IF: 3.8, Q2**.
8. Z Lu, **X Li**, M Rong, JS Baker, Y Gu. (2022). Effect of rearfoot valgus on biomechanics during barbell squatting: A study based on OpenSim musculoskeletal modeling. Frontiers in neurorobotics, 16, 832005. **IF: 2.8, Q2**.
9. Z Lu, **X Li**, R Xuan, Y Song, I Bíró, M Liang, Y Gu. (2022). Effect of heel lift insoles on lower extremity muscle activation and joint work during barbell squats. Bioengineering, 9(7), 301. **IF: 3.7, Q2**.
10. Z Lu, D Sun, D Xu, **X Li**, JS Baker, Y Gu. (2021). Gait characteristics and fatigue profiles when standing on surfaces with different hardness: Gait analysis and machine learning algorithms. Biology, 10(11), 1083. **IF: 3.5, Q2**.
11. L Shen, Z Lu, **X Li**, D Sun, Y Song, G Fekete, A Kovács, F Li, X Cen. Advances and Future Directions of Foetal Finite Element Modeling in Childbirth: from Biomechanical Interactions to Clinical Implications. International Journal of Biomedical Engineering and

Technology. **IF: 0.6, Q4**.

12. Chen, J., **Li, X.**, Hume, P. A., Wyatt, H., & Choisne, J. (2025). Multi-segment models for kinetic analysis of women during pregnancy: A systematic review. Gait & Posture, 2025. **IF: 2.4, Q2**.

International conference abstracts related to this thesis:

1. **Xin Li**, Lu Zhenghui, Gusztáv Fekete, András Kovács, Gu Yaodong The 17th Biennial Footwear Biomechanics Symposium. Title: Biomechanical Impact of Weight Gain and Ligament Relaxation on Arch Deformation During Pregnancy: A Finite Element Analysis. 2025. Norway (Oral presentation).
2. **Xin Li**, Lu Zhenghui, Gusztáv Fekete, András Kovács, Gu Yaodong. 2024 international Competitive Sports Biomechanics Forum 23rd National Sports Biomechanics Conference. Title: Analysis of Gait Characteristics During Pregnancy Based on Induced Acceleration. 2024. China (Oral presentation).
3. **Xin Li**, Lu Zhenghui, Gao Zixiang, Shen Siqin, He Yuqi, Gusztáv Fekete, András Kovács, Gu Yaodong. The 7th International Conference on Material Strength and Applied Mechanics (MSAM 2024). Title: The Impact of Shoe Heel-Toe Drop on Plantar Pressure during the Third Trimester of Pregnancy. 2024. Hungary (Oral presentation).

Other publications:

1. Lu, Z., **Li, X.**, Xuan, R., Song, Y., Bíró, I., Liang, M., & Gu, Y. (2022). Effect of heel lift insoles on lower extremity muscle activation and joint work during barbell squats. Bioengineering, 9(7), 301. **IF: 3.7, Q2**.
2. **Li X**, Adrien N, Baker J S, et al. Novice female exercisers exhibited different biomechanical loading profiles during full-squat and half-squat practice[J]. Biology, 2021, 10(11): 1184. **IF: 3.5, Q2**.
3. Lu, Z., Sun, D., Xu, D., **Li, X.**, Baker, J. S., & Gu, Y. (2021). Gait characteristics and fatigue profiles when standing on surfaces with different hardness: Gait analysis and machine learning algorithms. Biology, 10(11), 1083. **IF: 3.5, Q2**.
4. Xu Y, **Li X**, Sun Z, et al. Adjusted indirect and mixed comparisons of interventions for the Patient-Reported Outcomes Measures (PROMs) of disabled adults: A systematic review and network meta-analysis[J]. International Journal of Environmental Research and Public Health, 2021, 18(5): 2406. **IF: 2.92, Q1**

Reviewer for international journal articles:

1. Journal of Biomechanics
2. Footwear Science
3. Sport Sciences for Health
4. Physiology International
5. Physical Activity and Health Journal
6. International Journal of Environmental Research and Public Health
7. International Journal of Biomedical Engineering and Technology

ORCID: 0000-0003-1139-356X

Scopus Database: https://www.scopus.com/authid/detail.uri?authorId=57225159776

Independent citations: 96 (ResearchGate, December, 2025)

ResearchGate: https://www.researchgate.net/profile/Xin-Li-271?ev=hdr_xprf

Research Interest Score: 184.0 **(Web of Science)**

References

[1] A. L. Yaktine and K. M. Rasmussen, "Weight gain during pregnancy: reexamining the guidelines," 2010.

[2] L. A. Gilmore and L. M. Redman, "Weight gain in pregnancy and application of the 2009 IOM guidelines: toward a uniform approach," *Obesity,* vol. 23, no. 3, pp. 507-511, 2015.

[3] F. Hytten and G. Chamberlain, *Clinical physiology in obstetrics*. 1980.

[4] J. Most, K. L. Marlatt, A. D. Altazan, and L. M. Redman, "Advances in assessing body composition during pregnancy," *European journal of clinical nutrition,* vol. 72, no. 5, pp. 645-656, 2018.

[5] M. I. Ogamba, K. L. Loverro, N. M. Laudicina, S. V. Gill, and C. L. Lewis, "Changes in gait with anteriorly added mass: a pregnancy simulation study," *Journal of applied biomechanics,* vol. 32, no. 4, pp. 379-387, 2016.

[6] M. E. Franklin and T. Conner-Kerr, "An analysis of posture and back pain in the first and third trimesters of pregnancy," *Journal of Orthopaedic & Sports Physical Therapy,* vol. 28, no. 3, pp. 133-138, 1998.

[7] M. Betsch *et al.*, "Spinal posture and pelvic position during pregnancy: a prospective rasterstereographic pilot study," *European Spine Journal,* vol. 24, no. 6, pp. 1282-1288, 2015.

[8] R. Conder, R. Zamani, and M. Akrami, "The biomechanics of pregnancy: a systematic review," *Journal of functional morphology and kinesiology,* vol. 4, no. 4, p. 72, 2019.

[9] E. E. Butler, I. Colón, M. L. Druzin, and J. Rose, "Postural equilibrium during pregnancy: decreased stability with an increased reliance on visual cues," *American journal of obstetrics and gynecology,* vol. 195, no. 4, pp. 1104-1108, 2006.

[10] P. Ramachandra, P. Kumar, K. Bø, and G. A. Maiya, "Comparison of static postural sway characteristics between pregnant and non-pregnant women," *Journal of Biomechanics,* vol. 154, p. 111618, 2023.

[11] B. R. Dumke, L. H. Theilen, J. M. Shaw, K. B. Foreman, L. E. Dibble, and P. C. Fino, "Sensory integration and segmental control of posture during pregnancy," *Clinical Biomechanics,* vol. 115, p. 106264, 2024.

[12] L. S. Moreira, L. A. Elias, A. B. Gomide, M. F. Vieira, and W. N. Do Amaral, "A longitudinal assessment of myoelectric activity, postural sway, and low-back pain during pregnancy," *Acta of bioengineering and biomechanics,* vol. 19, no. 3, pp. 77-83, 2017.

[13] T. Foti, J. R. Davids, and A. Bagley, "A biomechanical analysis of gait during pregnancy," *JBJS,* vol. 82, no. 5, p. 625, 2000.

[14] J. J. Bagwell *et al.*, "Lower extremity kinetics and muscle activation during gait are significantly different during and after pregnancy compared to nulliparous females," *Gait & Posture,* vol. 81, pp. 33-40, 2020.

[15] J. S. Gottschall, R. C. Sheehan, and D. S. Downs, "Pregnant women exaggerate cautious gait patterns during the transition between level and hill surfaces," *Journal of electromyography and kinesiology,* vol. 23, no. 5, pp. 1237-1242, 2013.

[16] F. F. El-shamy, A. P. Ribeiro, and A. A. Abo Gazia, "Effectiveness of proprioceptive training on dynamic postural balance during pregnancy: A randomized controlled trial," *Physiotherapy Practice and Research,* vol. 40, no. 1, pp. 77-85, 2019.

[17] D. Aldabe, D. C. Ribeiro, S. Milosavljevic, and M. Dawn Bussey, "Pregnancy-related pelvic girdle pain and its relationship with relaxin levels during pregnancy: a systematic review," *European Spine Journal,* vol. 21, no. 9, pp. 1769-1776, 2012.

[18] C. Daneau, M. Houle, M. Pasquier, S.-M. Ruchat, and M. Descarreaux, "Association between pregnancy-related hormones and lumbopelvic pain characteristics in pregnant women: a scoping review," *Journal of manipulative and physiological therapeutics,* vol. 44, no. 7, pp. 573-583, 2021.

[19] F. Dehghan, B. Haerian, S. Muniandy, A. Yusof, J. Dragoo, and N. Salleh, "The effect of relaxin on the musculoskeletal system," *Scandinavian journal of medicine & science in sports,* vol. 24, no. 4, pp. e220-e229, 2014.

[20] T. Y. Chen *et al.*, "The relaxin family peptide receptor 1 (RXFP1): An emerging player in human health and disease," *Molecular Genetics & Genomic Medicine,* vol. 8, no. 4, p. e1194, 2020.

[21] H. Schock *et al.*, "Hormone concentrations throughout uncomplicated pregnancies: a longitudinal study," *BMC pregnancy and childbirth,* vol. 16, no. 1, p. 146, 2016.

[22] B. Yalçınkaya, E. A. Sezgin, K. G. Saçıntı, and L. Özçakar, "Neuromusculoskeletal disorders in pregnancy revisited: Insights and clinical implications," *Joint Diseases and Related Surgery,* vol. 36, no. 3, p. 741, 2025.

[23] Y. Cherni *et al.*, "Evaluation of ligament laxity during pregnancy," *Journal of gynecology obstetrics and human reproduction,* vol. 48, no. 5, pp. 351-357, 2019.

[24] M. Calguneri, H. Bird, and V. Wright, "Changes in joint laxity occurring during pregnancy," *Annals of the rheumatic diseases,* vol. 41, no. 2, pp. 126-128, 1982.

[25] C. W. Schauberger, B. L. Rooney, L. Goldsmith, D. Shenton, P. D. Silva, and A. Schaper, "Peripheral joint laxity increases in pregnancy but does not correlate with serum relaxin levels," *American journal of obstetrics and gynecology,* vol. 174, no. 2, pp. 667-671, 1996.

[26] M. L. Marnach, K. D. Ramin, P. S. Ramsey, S.-W. Song, J. J. Stensland, and K.-N. An, "Characterization of the relationship between joint laxity and maternal hormones in pregnancy," *Obstetrics & Gynecology,* vol. 101, no. 2, pp. 331-335, 2003.

[27] M. E. Bey, R. Marzilger, L. Hinkson, A. Arampatzis, and K. Legerlotz, "Patellar tendon stiffness is not reduced during pregnancy," *Frontiers in physiology,* vol. 10, p. 334, 2019.

[28] D. A. Hart, C. Reno, C. B. Frank, and N. G. Shrive, "Pregnancy affects cellular activity, but not tissue mechanical properties, in the healing rabbit medial collateral ligament," *Journal of Orthopaedic Research,* vol. 18, no. 3, pp. 462-471, 2000.

[29] Å. Rundgren, "Physical properties of connective tissue as influenced by single and repeated pregnancies in the rat," 1974.

[30] A. R. Bird, H. B. Menz, and C. C. Hyde, "The effect of pregnancy on footprint parameters. A prospective investigation," *Journal of the American Podiatric Medical Association,* vol. 89, no. 8, pp. 405-409, 1999.

[31] N. A. Segal, E. R. Boyer, P. Teran-Yengle, N. A. Glass, H. J. Hillstrom, and H. J. Yack, "Pregnancy leads to lasting changes in foot structure," *American journal of physical medicine & rehabilitation,* vol. 92, no. 3, pp. 232-240, 2013.

[32] M. O. B. Zangão, A. F. Poeira, M. Branco, and R. Santos-Rocha, "Changes in foot biomechanics during pregnancy and postpartum: Scoping review," *International journal of environmental research and public health,* vol. 21, no. 5, p. 638, 2024.

[33] A. Masłoń *et al.*, "Influence of pregnancy related anthropometric changes on plantar pressure distribution during gait—A follow-up study," *Plos one,* vol. 17, no. 3, p. e0264939, 2022.
[34] W.-K. Chiou, H.-T. Chiu, A.-S. Chao, M.-H. Wang, and Y.-L. Chen, "The influence of body mass on foot dimensions during pregnancy," *Applied ergonomics,* vol. 46, pp. 212-217, 2015.
[35] M. Alcahuz-Griñan, P. Nieto-Gil, P. Perez-Soriano, and G. Gijon-Nogueron, "Morphological and postural changes in the foot during pregnancy and puerperium: a longitudinal study," *International journal of environmental research and public health,* vol. 18, no. 5, p. 2423, 2021.
[36] S. M. Stearne, K. A. McDonald, J. A. Alderson, I. North, C. E. Oxnard, and J. Rubenson, "The foot's arch and the energetics of human locomotion," *Scientific reports,* vol. 6, no. 1, p. 19403, 2016.
[37] L. A. Kelly, D. J. Farris, A. G. Cresswell, and G. A. Lichtwark, "Intrinsic foot muscles contribute to elastic energy storage and return in the human foot," *Journal of Applied Physiology,* vol. 126, no. 1, pp. 231-238, 2019.
[38] K. A. Kirby, "Longitudinal arch load-sharing system of the foot," *Revista Española de Podología,* vol. 28, no. 1, pp. e18-e26, 2017.
[39] J. Bertuit, C. Leyh, M. Rooze, and V. Feipel, "Plantar pressure during gait in pregnant women," *Journal of the American Podiatric Medical Association,* vol. 106, no. 6, pp. 398-405, 2016.
[40] P. Ramachandra, P. Kumar, A. Kamath, and A. G. Maiya, "Do structural changes of the foot influence plantar pressure patterns during various stages of pregnancy and postpartum?," *Foot & ankle specialist,* vol. 10, no. 6, pp. 513-519, 2017.
[41] W. Forczek, Y. P. Ivanenko, J. Bielatowicz, and K. Wacławik, "Gait assessment of the expectant mothers–Systematic review," *Gait & posture,* vol. 62, pp. 7-19, 2018.
[42] C. Gaymer, H. Whalley, J. Achten, M. Vatish, and M. L. Costa, "Midfoot plantar pressure significantly increases during late gestation," *The Foot,* vol. 19, no. 2, pp. 114-116, 2009.
[43] E. Karadag-Saygi, F. Unlu-Ozkan, and A. Basgul, "Plantar pressure and foot pain in the last trimester of pregnancy," *Foot & ankle international,* vol. 31, no. 2, pp. 153-157, 2010.
[44] W. Forczek and R. Staszkiewicz, "Changes of kinematic gait parameters due to pregnancy," *Acta Bioeng Biomech,* vol. 14, no. 4, pp. 113-119, 2012.
[45] M. Branco, R. Santos-Rocha, L. Aguiar, F. Vieira, and A. Veloso, "Kinematic analysis of gait in the second and third trimesters of pregnancy," *Journal of pregnancy,* vol. 2013, no. 1, p. 718095, 2013.
[46] J. W. Błaszczyk, A. Opala-Berdzik, and M. Plewa, "Adaptive changes in spatiotemporal gait characteristics in women during pregnancy," *Gait & posture,* vol. 43, pp. 160-164, 2016.
[47] Z. Krkeljas, "Changes in gait and posture as factors of dynamic stability during walking in pregnancy," *Human movement science,* vol. 58, pp. 315-320, 2018.
[48] A. M. ElDeeb, H. A. Hamada, A. A. Abdel-Aziem, and A. M. Youssef, "The relationship between trunk and pelvis kinematics during pregnancy trimesters," *Acta of Bioengineering and Biomechanics,* vol. 18, no. 4, 2016.
[49] W. L. Gilleard, "Trunk motion and gait characteristics of pregnant women when walking: report of a longitudinal study with a control group," *BMC pregnancy and childbirth,* vol.

13, no. 1, p. 71, 2013.
[50] J. J. Bagwell *et al.*, "An exploratory analysis of gait biomechanics and muscle activation in pregnant females with high and low scores for low back or pelvic girdle pain during and after pregnancy," *Clinical Biomechanics,* vol. 97, p. 105705, 2022.
[51] Q. Mei, Y. Gu, and J. Fernandez, "Alterations of pregnant gait during pregnancy and post-partum," *Scientific reports,* vol. 8, no. 1, p. 2217, 2018.
[52] L. Aguiar, R. Santos-Rocha, M. Branco, F. Vieira, and A. Veloso, "Biomechanical model for kinetic and kinematic description of gait during second trimester of pregnancy to study the effects of biomechanical load on the musculoskeletal system," *Journal of mechanics in medicine and biology,* vol. 14, no. 01, p. 1450004, 2014.
[53] A. Haddox, J. Hausselle, and A. Azoug, "Changes in segmental mass and inertia during pregnancy: A musculoskeletal model of the pregnant woman," *Gait & Posture,* vol. 76, pp. 389-395, 2020.
[54] S. Morino, M. Yamashita, F. Umezaki, H. Hatanaka, and M. Takahashi, "Assessment of motion and muscle activation impacts on low back pain during pregnancy using an inertial measurement unit," *Applied Sciences,* vol. 10, no. 11, p. 3690, 2020.
[55] J. Chen, X. Li, P. A. Hume, H. Wyatt, and J. Choisne, "Multi-segment models for kinetic analysis of women during pregnancy: A systematic review," *Gait & Posture,* 2025.
[56] X. Li *et al.*, "Effect of pregnancy on female gait characteristics: A pilot study based on portable gait analyzer and induced acceleration analysis," *Frontiers in Physiology,* vol. 14, p. 1034132, 2023.
[57] S. Morino and M. Takahashi, "Estimating Co-Contraction activation of trunk muscles using a Novel musculoskeletal model for pregnant women," *Applied Sciences,* vol. 7, no. 10, p. 1067, 2017.
[58] K. Dunning, G. LeMasters, and A. Bhattacharya, "A major public health issue: the high incidence of falls during pregnancy," *Maternal and child health journal,* vol. 14, no. 5, pp. 720-725, 2010.
[59] J. O. Awoleke, B. A. Olofinbiyi, A. O. Awoleke, and A. C. Omoyajowo, "Obstetric correlates of maternal falls in Southern Nigeria," *The Scientific World Journal,* vol. 2019, no. 1, p. 9716919, 2019.
[60] J. McCrory, A. Chambers, A. Daftary, and M. Redfern, "Dynamic postural stability during advancing pregnancy," *Journal of biomechanics,* vol. 43, no. 12, pp. 2434-2439, 2010.
[61] A. Inanir, B. Cakmak, Y. Hisim, and F. Demirturk, "Evaluation of postural equilibrium and fall risk during pregnancy," *Gait & posture,* vol. 39, no. 4, pp. 1122-1125, 2014.
[62] A. Opala-Berdzik *et al.*, "Static postural stability in women during and after pregnancy: a prospective longitudinal study," *PloS one,* vol. 10, no. 6, p. e0124207, 2015.
[63] J. J. Bagwell *et al.*, "Center of pressure characteristics differ during single leg stance throughout pregnancy and compared to Nulligravida individuals," *Gait & Posture,* vol. 97, pp. 43-47, 2022.
[64] L. Wei *et al.*, "Suitable heel height, a potential method for musculoskeletal problems during the third trimester: a pilot study," *Bioengineering,* vol. 11, no. 7, p. 667, 2024.
[65] S. I. Jang, Y. R. Lee, H. S. Kwak, K. S. Moon, J.-C. Shin, and J.-H. Kim, "The effect of balanced incline shoes on walking and feet for the pregnant women," *Korean Journal of Obstetrics and Gynecology,* vol. 53, no. 11, pp. 988-997, 2010.

[66] A. Marques, P. Gonçalves, R. Santos, and J. Vilas Boas, "Comfort and functionality of pregnant women's feet study of kinetic parameters with silicon insoles," *Brazilian Journal of Biomechanics,* vol. 6, no. 10, pp. 9-15, 2005.

[67] N. A. Segal, L. N. Neuman, M. C. Hochstedler, and H. L. Hillstrom, "Static and dynamic effects of customized insoles on attenuating arch collapse with pregnancy: A randomized controlled trial," *The Foot,* vol. 37, pp. 16-22, 2018.

[68] M. Gimunova, M. Zvonar, M. Sebera, P. Turcinek, and K. Kolarova, "Special footwear designed for pregnant women and its effect on kinematic gait parameters during pregnancy and postpartum period," *Plos One,* Article vol. 15, no. 5, May 12 2020, Art. no. e0232901.

[69] R. Díaz-Meco Conde *et al.*, "Influence of unstable shoes on women with lumbopelvic postpartum pain: randomized clinical trial," *São Paulo medical journal,* vol. 139, pp. 312-318, 2021.

[70] F. Faul, E. Erdfelder, A.-G. Lang, and A. Buchner, "G* Power 3: A flexible statistical power analysis program for the social, behavioral, and biomedical sciences," *Behavior research methods,* vol. 39, no. 2, pp. 175-191, 2007.

[71] Z. Lu *et al.*, "Parametric cushioning lattice insole based on finite element method and machine learning: A preliminary computational analysis," *Journal of Biomechanics,* vol. 184, p. 112674, 2025.

[72] M. H. Ramlee, M. R. A. Kadir, M. R. Murali, and T. Kamarul, "Biomechanical evaluation of two commonly used external fixators in the treatment of open subtalar dislocation—A finite element analysis," *Medical Engineering & Physics,* vol. 36, no. 10, pp. 1358-1366, 2014.

[73] P. Golanó *et al.*, "Anatomy of the ankle ligaments: a pictorial essay," *Knee Surgery, Sports Traumatology, Arthroscopy,* vol. 24, pp. 944-956, 2016.

[74] T. L.-W. Chen, Y. Wang, Y. Peng, G. Zhang, T. T.-H. Hong, and M. Zhang, "Dynamic finite element analyses to compare the influences of customised total talar replacement and total ankle arthroplasty on foot biomechanics during gait," *Journal of orthopaedic translation,* vol. 38, pp. 32-43, 2023.

[75] T. L.-W. Chen, D. W.-C. Wong, Y. Peng, and M. Zhang, "Prediction on the plantar fascia strain offload upon Fascia taping and Low-Dye taping during running," *Journal of orthopaedic translation,* vol. 20, pp. 113-121, 2020.

[76] A. Perrier, V. Luboz, M. Bucki, F. Cannard, N. Vuillerme, and Y. Payan, "Biomechanical modeling of the foot," in *Biomechanics of Living Organs*: Elsevier, 2017, pp. 545-563.

[77] C.-Y. Hsu, C.-S. Wang, K.-W. Lin, M.-J. Chien, S.-H. Wei, and C.-S. Chen, "Biomechanical analysis of the flatfoot with different 3D-printed insoles on the lower extremities," *Bioengineering,* vol. 9, no. 10, p. 563, 2022.

[78] B. Nie *et al.*, "Determination of the in situ mechanical behavior of ankle ligaments," *Journal of the mechanical behavior of biomedical materials,* vol. 65, pp. 502-512, 2017.

[79] S. Siegler, J. Block, and C. D. Schneck, "The mechanical characteristics of the collateral ligaments of the human ankle joint," *Foot & ankle,* vol. 8, no. 5, pp. 234-242, 1988.

[80] N. A. Maffiuletti *et al.*, "Concurrent validity and intrasession reliability of the IDEEA accelerometry system for the quantification of spatiotemporal gait parameters," *Gait & posture,* vol. 27, no. 1, pp. 160-163, 2008.

[81] A. Yawar, L. Korpas, S. Mandre, and M. Venkadesan, "Transverse contributions to the longitudinal stiffness of the human foot," *arXiv preprint arXiv:1706.04610,* 2017.

[82] E. Schmidt *et al.*, "The role of the transverse arch in progressive collapsing foot deformity," *Foot & Ankle International,* vol. 45, no. 1, pp. 44-51, 2024.

[83] M. M. Simonik, J. Pitarresi, and R. Willing, "Analysis of intramedullary beam designs using customized finite element models for medial column arthrodesis of the foot," *The Journal of Foot and Ankle Surgery,* vol. 61, no. 3, pp. 508-519, 2022.

[84] L. Gao, Z. Lu, M. Liang, J. S. Baker, and Y. Gu, "Influence of Different Load Conditions on Lower Extremity Biomechanics during the Lunge Squat in Novice Men," *Bioengineering,* vol. 9, no. 7, p. 272, 2022.

[85] Z. Lu, X. Li, M. Rong, J. S. Baker, and Y. Gu, "Effect of rearfoot valgus on biomechanics during barbell squatting: A study based on OpenSim musculoskeletal modeling," *Frontiers in Neurorobotics,* vol. 16, 2022.

[86] Z. Lu *et al.*, "Effect of Heel Lift Insoles on Lower Extremity Muscle Activation and Joint Work during Barbell Squats," *Bioengineering,* vol. 9, no. 7, p. 301, 2022.

[87] G. Chen, "Induced acceleration contributions to locomotion dynamics are not physically well defined," *Gait & posture,* vol. 23, pp. 37-44, 2006.

[88] T. R. Souza *et al.*, "Muscle actions on crossed and non-crossed joints during upright standing and gait: A comprehensive description based on induced acceleration analysis," *Journal of biomechanics,* vol. 130, p. 110874, 2022.

[89] F. E. Zajac and M. E. Gordon, "Determining muscle's force and action in multi-articular movement," *Exercise and sport sciences reviews,* vol. 17, pp. 187-230, 1989.

[90] S. R. Hamner and S. L. Delp, "Muscle contributions to fore-aft and vertical body mass center accelerations over a range of running speeds," *Journal of biomechanics,* vol. 46, no. 4, pp. 780-787, 2013.

[91] S. R. Hamner, A. Seth, and S. L. Delp, "Muscle contributions to propulsion and support during running," *Journal of biomechanics,* vol. 43, no. 14, pp. 2709-2716, 2010.

[92] S. R. Hamner, A. Seth, K. M. Steele, and S. L. Delp, "A rolling constraint reproduces ground reaction forces and moments in dynamic simulations of walking, running, and crouch gait," *Journal of biomechanics,* vol. 46, no. 10, pp. 1772-1776, 2013.

[93] R. Ker, M. Bennett, S. Bibby, R. Kester, and R. M. Alexander, "The spring in the arch of the human foot," *Nature,* vol. 325, no. 6100, pp. 147-149, 1987.

[94] L. Zhang, Q. Zhang, Y. Zhong, T. Hortobagyi, and Y. Gu, "Effect of forefoot transverse arch stiffness on foot biomechanical response--based on finite element method," *Frontiers in Bioengineering and Biotechnology,* vol. 12, p. 1387768, 2024.

[95] K. J. Mickle, B. J. Munro, S. R. Lord, H. B. Menz, and J. R. Steele, "Foot pain, plantar pressures, and falls in older people: a prospective study," *Journal of the American Geriatrics Society,* vol. 58, no. 10, pp. 1936-1940, 2010.

[96] M. Venkadesan *et al.*, "Stiffness of the human foot and evolution of the transverse arch," *Nature,* vol. 579, no. 7797, pp. 97-100, 2020.

[97] L. Krüger *et al.*, "Influence of the transverse tarsal arch on radiological components of progressive collapsing foot deformity," *Journal of Orthopaedic Research®,* vol. 42, no. 12, pp. 2752-2760, 2024.

[98] H. Zeidan *et al.*, "Detailed analysis of the transverse arch of hallux valgus feet with and

without pain using weightbearing ultrasound imaging and precise force sensors," *PLoS One,* vol. 15, no. 1, p. e0226914, 2020.

[99] A. Asghar and S. Naaz, "The transverse arch in the human feet: A narrative review of its evolution, anatomy, biomechanics and clinical implications," *Morphologie,* vol. 106, no. 355, pp. 225-234, 2022.

[100] A. Hills, E. Hennig, M. McDonald, and O. Bar-Or, "Plantar pressure differences between obese and non-obese adults: a biomechanical analysis," *International journal of obesity,* vol. 25, no. 11, pp. 1674-1679, 2001.

[101] L. A. Kelly, A. G. Cresswell, S. Racinais, R. Whiteley, and G. Lichtwark, "Intrinsic foot muscles have the capacity to control deformation," 2014.

[102] P. A. Gribble and J. Hertel, "Effect of lower-extremity muscle fatigue on postural control," *Archives of physical medicine and rehabilitation,* vol. 85, no. 4, pp. 589-592, 2004.

[103] W. L. Gilleard, "Trunk motion and gait characteristics of pregnant women when walking: report of a longitudinal study with a control group," *BMC pregnancy and childbirth,* vol. 13, pp. 1-8, 2013.

[104] H. Yoo, D. Shin, and C. Song, "Changes in the spinal curvature, degree of pain, balance ability, and gait ability according to pregnancy period in pregnant and nonpregnant women," *Journal of physical therapy science,* vol. 27, pp. 279-284, 2015.

[105] J. W. Błaszczyk, M. Plewa, J. Cieślinska-Swider, B. Bacik, B. Zahorska-Markiewicz, and A. Markiewicz, "Impact of excess body weight on walking at the preferred speed," *Acta neurobiologiae experimentalis,* vol. 71, pp. 528-540, 2011.

[106] S. Wearing, E. Hennig, N. Byrne, J. Steele, and A. Hills, "The biomechanics of restricted movement in adult obesity," *Obesity reviews,* vol. 7, pp. 13-24, 2006.

[107] A. P. Ribeiro, S. M. A. João, and I. C. N. Sacco, "Static and dynamic biomechanical adaptations of the lower limbs and gait pattern changes during pregnancy," *Women's Health,* vol. 9, pp. 99-108, 2013.

[108] L. E. Nagy and J. C. King, "Energy expenditure of pregnant women at rest or walking self-paced," *The American journal of clinical nutrition,* vol. 38, pp. 369-376, 1983.

ACKNOWLEDGEMENTS

Reaching the culmination of my doctoral journey is a milestone made possible by the invaluable contributions of many individuals and institutions. My most profound appreciation is reserved for my supervisors: Dr. habil. Gusztáv Fekete, Dr. András Kovács, and Prof. Dr. Yaodong Gu. Their mentorship has been characterized by critical insight, patient instruction, and constant encouragement throughout this project. This thesis has been significantly shaped by their scholarly wisdom and expertise.

I also wish to express my gratitude to the Faculty of Engineering at the University of Pannonia (PE), the Savaria Institute of Technology at Eötvös Loránd University (ELTE), and the Faculty of Sports Science at Ningbo University (NBU). The essential infrastructure and academic resources provided by these esteemed organizations laid the groundwork for my experiments and analysis.

Furthermore, I am indebted to my respected mentors and peers, specifically Prof. Dr. István Bíró, Dr. Ildikó Molnár, Dr. Endre Jánosi, and Dr. Tej Singh, alongside my fellow graduate students. I sincerely appreciate your diverse assistance, which spanned from academic discussions to personal support. Crucially, this research fundamentally relied on the voluntary involvement of the study participants; their time and effort were vital to the data collection process.

On a personal level, my journey has been sustained by the unwavering backing of my family. I am eternally grateful to my parents, Mr. Li Yimin and Mrs. Liu Xiangchen, for their steadfast belief in my potential. To my fiancé, Mr. Lu Zhenghui, thank you for serving as my pillar of strength; your unconditional love and motivation have kept me going.

Finally, I acknowledge the critical funding provided by the Stipendium Hungaricum Programme, the Tempus Public Foundation, and the China Scholarship Council (CSC), without which this research endeavor would not have been feasible.