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Biomechanical Evaluation of Foot Orthoses for Flatfoot in Diverse Movement Tasks and Exploration of Digital Design Strategies

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Motivation of the work

Flatfoot, also referred to as pes planus, is one of the most common deformities of the lower limb[1]. It is characterized primarily by collapse of the medial longitudinal arch, accompanied by excessive calcaneal eversion and forefoot pronation. Such abnormal foot postures are often associated with structural changes such as ligamentous laxity, elongation of the plantar fascia, and weakness of the tibialis posterior muscle, and they in turn alter the movement patterns of proximal joints such as the ankle and knee, increasing plantar loading in specific regions[2]. These alterations may disrupt normal shock absorption mechanisms and increase the risk of overuse injuries such as achilles tendinopathy, tibial stress fractures, and plantar fasciitis[3]. For patients with flatfoot, non-invasive interventions such as foot orthoses (FOs) are commonly used in clinical practice to restore foot alignment and improve movement comfort by supporting the arch, redistributing plantar pressure, and limiting excessive pronation[4]. Existing studies indicate that FOs optimize foot alignment and protect the medial longitudinal arch from abnormal stress, thereby enhancing arch-support function and stability[5,6]. Despite their widespread clinical application, the biomechanical mechanisms by which foot orthosis (FO) design features act in individuals with flatfoot remain insufficiently understood, and the interactions between FOs and different movement tasks (such as walking, running, and landing) are not comprehensively evaluated[7,8]. These gaps limit the evidence-based selection and optimization of FO prescriptions as well as the development of rational design strategies.

One of the core parameters in FO design is arch-support hardness. The hardness of material is directly related to functional outcomes and symptom relief in the treatment of flatfoot. When hardness is inappropriate, the FOs cannot effectively correct arch collapse or control excessive pronation, and thus cannot adequately relieve compensatory muscle tension, gait instability, and reduced gait efficiency[6,9,10]. High-hardness materials can restore the height of the medial arch to a greater extent and better resist deformation under body weight, thereby preventing arch collapse; however, this enhanced support also

imposes “unreasonable” increases in pressure on the talocalcaneal joint, plantar ligaments, and other midfoot structures[11,12]. These findings highlight the inevitable trade-offs involved in determining material hardness. In clinical practice, clinicians often rely on subjective experience to adjust arch-support hardness in an attempt to balance comfort, support, and motion control[13]. Most existing studies focus only on changes in single outcome measures and seldom employ a unified experimental framework to examine how FOs with different arch-support hardnesses affect multiple dimensions—such as lower-limb joint kinematics, joint moments, and joint reaction forces—throughout gait[11,14,15]. In addition, traditional statistical methods are commonly used, making it difficult to fully characterize uncertainty in the results[16]. Therefore, more comprehensive biomechanical research and advanced analytical methods are needed to systematically assess how arch-support hardness influences lower-limb mechanical responses in individuals with flexible flatfoot, so as to support the optimization of clinical FO design and prescription.

Compared with walking, running is more demanding in terms of ground reaction forces, loading rates, and thus represents a particularly critical locomotor task in flatfoot management[17]. Many runners exhibit flatfoot characteristics and frequently use FOs while running to mitigate excessive pronation, reduce impact loading, and improve comfort and performance[18]. Under the high repetition, high impact conditions of running, FOs not only influence foot posture but may also alter shock absorption, energy transfer, and the coordination patterns of proximal joints, including the ankle, knee, and hip, throughout the stance phase[18]. At the same time, kinesio tape (KT) is widely used as an alternative or adjunctive intervention to support the arch and regulate foot motion with minimal changes to footwear conditions[19]. However, systematic research directly quantifying the differences in biomechanical effects between FOs and KT during running in individuals with flatfoot remains limited, especially studies that simultaneously consider multiple dimensions such as three-dimensional kinematics, kinetics, and joint coordination variability[20]. A more comprehensive understanding of how FOs modify running biomechanics, and how these effects compare with tape interventions, is crucial

for clarifying the relative advantages and limitations of FOs and for guiding the selection of appropriate interventions for runners with flatfoot.

In addition to walking and running, many daily and sporting situations involve landing tasks from different heights, such as stair descent and drop jumps[21]. These tasks generate higher ground impact forces and loading rates, which may exacerbate the biomechanical disadvantages associated with flatfoot and increase the risk of injuries to the foot and ankle as well as the knee and hip joints[22]. Research have showed that the foot arch deforms markedly during landing, and that individuals with flexible flatfoot demonstrate significantly improved plantar pressure distribution after continuous FO use; this finding supports the beneficial role of FOs intervention for landing in flatfoot populations[1,23]. Clinically, the use of FOs is also often recommended in impact tasks to provide additional support and cushioning[24]. However, evidence on how FOs modulate impact and lower-limb joint biomechanics across different landing heights remains limited, highlighting the need for comprehensive investigations to clarify their applicability and potential protective role in high impact movements.

While experimental studies investigate the biomechanical effects of FOs, the digitalization of individualized FO design and manufacturing processes is also an urgent issue to be addressed[25]. Traditional FO fabrication relies heavily on the experience and manual craftsmanship of physicians and orthotists, often requiring multiple fitting and adjustment sessions; this process is time-consuming, difficult to standardize, and challenging to scale up[26]. In recent years, the development of three-dimensional scanning, finite element analysis (FEA), and additive manufacturing (3D printing) technologies makes it feasible to integrate fully digital design workflows[27,28]. By combining finite element analysis with machine learning to construct a digital workflow for personalized 3D printed insoles, it becomes possible to provide a technical foundation for future FO designs targeting various foot problems, including flatfoot.

Research objectives

Based on the work motivation, the main research objectives of this dissertation are as follows.

The first research objective: To investigate the biomechanical effects of foot orthoses with different arch support hardnesses on the hip, knee, ankle, and foot joints in adults with flatfoot during the stance phase of walking, as well as their interrelationships, including joint angles, moments, joint reaction forces, and muscle forces, through musculoskeletal simulation combined with Bayesian analysis. This aimed to reveal how biomechanical benefits varied with hardness, evaluated appropriate arch support hardness, and provided a basis for optimizing arch support hardness design. It was hypothesized that: (1) as hardness increased, peak tibialis posterior muscle force further decreased; (2) knee joint load-related parameters increased with increasing hardness; (3) ankle and foot biomechanical measures showed further improvement as hardness increased.

The second research objective: To systematically analyze the effects of foot orthoses on runners with flatfoot during the running stance phase across multiple biomechanical levels using musculoskeletal modeling methods (assessing angles and moments), vector coding technique (assessing inter-joint coordination and variability), continuous wavelet transform (analyzing time-frequency characteristics of ground reaction forces), and finite element analysis (calculating stress on foot bones and fascia). This was compared with the effects of kinesio tape intervention to determine which more effectively corrected poor biomechanics and reduced potential injury risks, thereby revealing the advantages and limitations of foot orthoses during running and providing scientific evidence for clinical selection.

The third research objective: To investigate the effects of foot orthoses on landing biomechanics in individuals with flatfoot under different landing height conditions. By comparing key indicators (lower limb joint angles, joint moments, and ground reaction forces) with and without foot orthoses at various landing heights, this research aimed to clarify whether the intervention effect remained stable as landing impact intensity

increased and elucidated how foot orthoses modulated movement patterns and load distribution during landing in individuals with flatfoot.

The fourth research objective: To construct an integrated workflow, taking foot orthoses for diabetic foot—which was sensitive to plantar pressure distribution and had the lowest design tolerance—as an example: generating variants of foot support structures through parametric computer-aided design, systematically evaluating the impact of different structures on pressure distribution using finite element analysis, and training a convolutional neural network algorithm based on simulation data to establish a personalized insole structure and material thickness plan based on ground reaction force output. This integrated workflow was proposed to provide an efficient engineering methodology for the development of personalized foot orthoses and could be extended for flatfoot.

1 LITERATURE REVIEW

1.1 Biomechanics of Flatfoot

1.1.1 Definition, Classification, and Epidemiology

Flatfoot (pes planus) was not merely a “lowered arch”, but a complex multi-planar deformity involving coupled malalignments across the rearfoot, midfoot, and forefoot. The hallmark structural feature was the collapse of the medial longitudinal arch (MLA), but the clinical and biomechanical presentation typically included excessive subtalar pronation, calcaneal eversion (hindfoot valgus), and forefoot abduction relative to the hindfoot, often producing the characteristic “too-many-toes” sign in more pronounced cases (**Figure 1**)[29]. Under weight-bearing, the talus tended to plantarflex and adduct, shifting its position relative to the calcaneus and altering midfoot congruency. This change “unlocked” the midtarsal joints and reduced the stability of the osseous “truss” making the foot more compliant when it should have function as a stable base[29].



Figure 1 Differences between flatfoot and normal foot

Clinically, flatfoot was commonly classified into flexible and rigid forms. Flexible flatfoot—the most prevalent form in both pediatric and adult populations—was defined by an MLA that appeared in non-weight-bearing or during a single-leg heel rise, but collapsed during stance due to ligamentous laxity, altered neuromuscular control, and insufficient dynamic support[30]. In contrast, rigid flatfoot presented with a persistently absent arch regardless of loading, frequently reflecting structural constraints or bony pathology (e.g., tarsal coalition), and is more often associated with limited subtalar motion and pain[30]. Adult-acquired flatfoot deformity (AAFD) was frequently

conceptualized as a progressive pathway in which a previously flexible or compensated foot became increasingly decompensated, often involving posterior tibial tendon dysfunction and progressive hindfoot valgus[31].

From an epidemiological perspective, prevalence estimates varied with definition, measurement method, and population, but available data indicated that flatfoot was common. Adult prevalence was reported in the range of approximately 13.6% to 27%, and pediatric prevalence could be substantially higher (reported up to approximately 44% in preschool-aged children), with many pediatric cases improving as tissues matured and neuromuscular control developed[32]. This wide range underscores that flatfoot was better understood as a spectrum rather than a single entity, and clinical relevance depended on symptoms, function, and task demands, not simply static morphology. Although many individuals remained asymptomatic, the altered load distribution and reduced mechanical efficiency associated with persistent arch collapse could increase susceptibility to fatigue, local tissue overload, and proximal compensations, particularly during high-demand activities that amplify internal joint moments and soft-tissue strain[29,30,32].

1.1.2 Intra-pedal Biomechanics: The Failed Rigid Lever

A central functional deficit in symptomatic pes planus was the inability of the foot to perform its normal phase-dependent transformation during gait, shifting from a compliant “mobile adaptor” at early stance to a stable “rigid lever” at terminal stance. In typical gait, pronation during loading response was beneficial because it increased contact area, supported shock attenuation, and helped the foot adapt to surface irregularities. As the body progresses over the foot, effective propulsion required resupination, which tightened joint coupling, aligned the midfoot, and locked themidtarsal joints (talonavicular and calcaneocuboid) to provide a stable lever for push-off[33].

In flexible flatfoot, this transition was often delayed, incomplete, or absent. The rearfoot remained excessively everted and the midfoot remained relatively “unlocked,” which compromised the rigidity required for efficient force transfer. Functionally, this meant that during late stance, when the plantarflexors and intrinsic foot muscles should

have leveraged a stable arch to generate forward propulsion, the foot continued to deform, absorbing energy that would otherwise contribute to propulsion. This could be conceptualized as an energy-loss mechanism, in which mechanical work was dissipated through excessive joint motion and soft-tissue deformation rather than stored and returned effectively.

The Windlass Mechanism, classically described by Hicks, was critical to late-stance stiffening (**Figure 2**). During terminal stance, dorsiflexion of the hallux tensioned the plantar fascia, elevated the MLA, and contributed to a stiffer midfoot for toe-off[34]. In a flatfoot, persistent arch collapse and altered midfoot alignment could introduce slack into the plantar fascia system, diminishing the effective windlass tensioning and reducing the rise of the MLA that normally assists rigid lever formation[34]. These deficiencies could be more consequential in dynamic tasks such as running and landing, where joint moments and plantarflexion demands were higher, and the need for rapid stiffness modulation became greater.

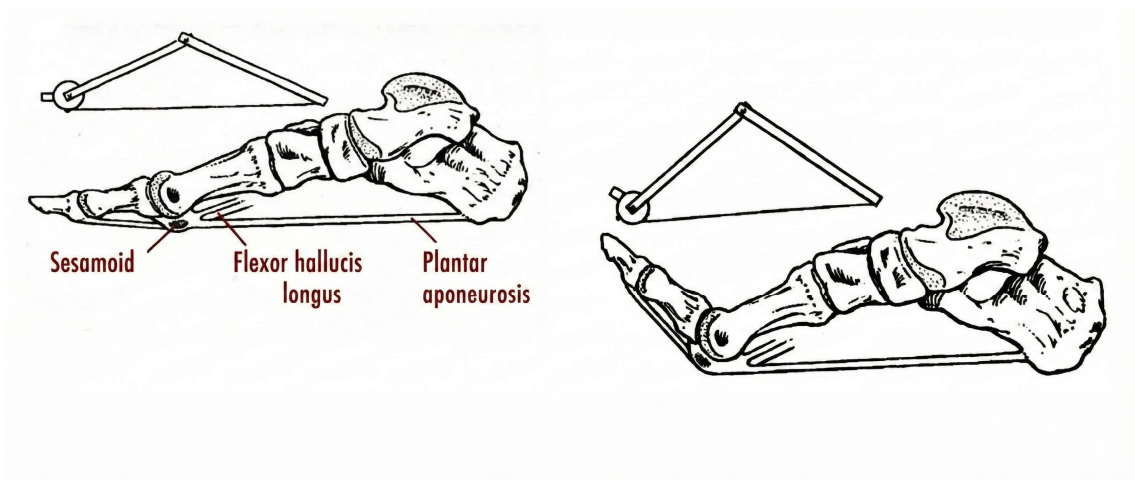


Figure 2 Foot windlass mechanism

Kinetically, gait analyses in flatfoot populations frequently showed alterations in center of pressure (COP) progression and vertical ground reaction force (vGRF) patterns, reflecting changes in loading strategy and propulsive efficiency. A more medial COP trajectory was often interpreted as evidence of increased midfoot and medial column loading and prolonged pronation, while changes in vGRF peaks might have indicated

differences in impact attenuation and push-off mechanics[35]. Over time, these deviations could increase muscular demand because extrinsic muscles (e.g., tibialis posterior) and intrinsic foot muscles had to work harder to stabilize a hypermobile structure, raising the energetic cost of walking and contributing to activity-related fatigue commonly reported by symptomatic individuals[35].

1.1.3 Soft Tissue Strain and Pathologies

The structural and kinematic features of pes planus imposed chronic, repetitive strain on the soft tissues responsible for arch integrity. The plantar fascia was a major passive stabilizer of the MLA; when the arch collapsed, the fascia experienced increased tensile loading, prolonged elongation, and repetitive microtrauma. This provided a plausible biomechanical explanation for the strong clinical association between flatfoot and plantar fasciitis and fasciopathy, where pain often localized near the medial calcaneal tubercle[36]. While symptoms are multifactorial, the mechanical environment of a flattened arch could foster degenerative change through repeated strain under load and impaired energy return during gait.

Imaging findings supported this tissue-overload model. Ultrasonography commonly revealed increased plantar fascia thickness, particularly near the calcaneal insertion, consistent with chronic degeneration rather than purely acute inflammation[37]. For orthoses research, these structural changes indicated that the target of intervention was not only “raising an arch”, but also modifying tissue loading by reducing peak strain, altering the timing of load, and improving late-stance stiffness so the plantar fascia was not required to compensate for persistent instability[36,37].

In addition, the posterior tibialis tendon (PTT) was a key dynamic stabilizer of the arch and hindfoot (**Figure 3**). The tibialis posterior contributed to inversion and plantarflexion moments that resist excessive pronation and support midfoot stability. In flatfoot mechanics, the PTT was often forced into sustained and sometimes eccentrically dominant activity to oppose calcaneal eversion torque and midfoot collapse. Over time, the tendon might have undergone overload-related degeneration and elongation, leading

to posterior tibial tendon dysfunction (PTTD), a commonly described pathway in adult-acquired flatfoot deformity[31]. As PTT competence declined, an essential dynamic constraint was weakened, increasing the likelihood of deformity progression and conversion from a flexible presentation toward a more rigid and symptomatic condition[31].

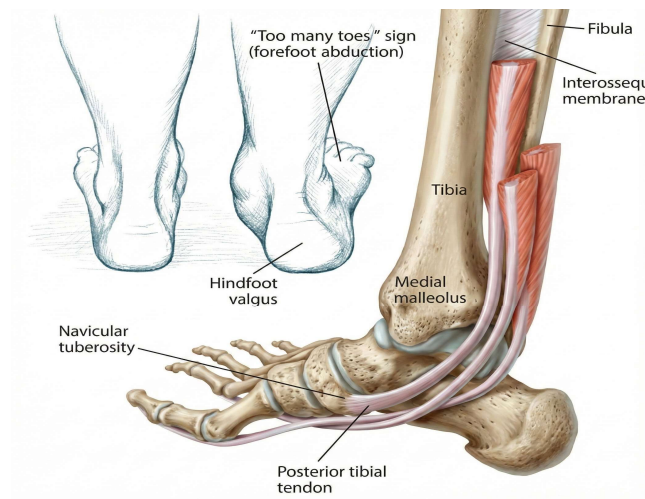


Figure 3 PTTD: Anatomy and Clinical Presentation

Overall, pes planus could be reframed as a load-management problem. When passive and active stabilizers (plantar fascia, PTT, spring ligament complex) were repeatedly overloaded, tissue capacity could be exceeded, and compensation strategies became less effective. This provided a mechanistic rationale for interventions combining mechanical support (orthoses) and capacity-building (strengthening and neuromuscular training), aiming to reduce pathological strain while improving the foot’s ability to generate and regulate stiffness under functional demands[31,36,37].

1.1.4 Proximal Kinetic Chain Effects

Pes planus might have influenced the entire lower extremity kinetic chain because subtalar mechanics were coupled with tibial rotation. The subtalar joint was described as functioning like a “miter gear”, whereby frontal-plane motion of the rearfoot (eversion or inversion) was mechanically coupled with transverse-plane rotation of the tibia (**Figure 4**)[38]. When subtalar pronation was excessive or prolonged, the tibia tended to undergo

greater magnitude and longer-duration internal rotation, altering the kinematic environment at the knee and hip[38].

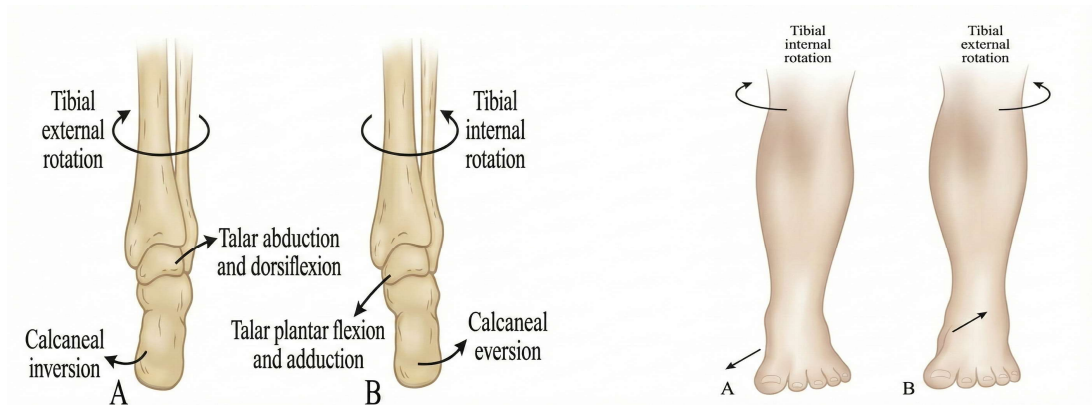


Figure 4 Coupled motions of tibial rotation and subtalar joint kinematics

At the knee, increased tibial internal rotation could have contributed to a dynamic valgus vector when combined with hip adduction or internal rotation and insufficient proximal control. Even if the relationship was not deterministic for every individual, this coupling provided a plausible biomechanical pathway linking distal alignment to knee joint orientation and potentially increased stress on passive restraints and patellofemoral structures[38].

With respect to patellofemoral symptoms, increased tibial internal rotation could modify the effective alignment of the extensor mechanism and influence patellar tracking. When the tibia rotated internally relative to the femur, the relative lateral force vector acting on the patella might have increased, which could have contributed to maltracking and elevated contact pressures at the patellofemoral joint. These mechanisms were frequently discussed in relation to patellofemoral pain syndrome (PFPS) and might have helped explain why some individuals experienced anterior knee pain alongside foot pronation abnormalities[39].

Over longer time scales, altered alignment and rotational mechanics might also have influenced cartilage loading patterns and shear forces at the tibiofemoral joint. Although knee osteoarthritis was multifactorial, persistent malalignment and abnormal rotation could change the distribution of loads across compartments and might have accelerated

degenerative processes in susceptible individuals[40]. From a biomechanical perspective, the key implication was that pes planus might have altered not only plantar load distribution but also tibial–femoral rotational dynamics, linking orthotic intervention to clinically meaningful outcomes beyond the foot[38-40].

At the hip, effects were more variable and strongly mediated by proximal strength and motor control; however, distal alignment could still have shaped proximal kinematics, especially during high-load tasks (running and landing) where joint coupling and neuromuscular demands intensify. This supported evaluating orthoses across diverse movement tasks, since compensations might have been minimal in level walking but became evident under increased mechanical demand.

1.1.5 Clinical Management

A mechanistic understanding of pes planus supported treatment strategies that aimed to reduce excessive pronation and midfoot collapse when it was functionally problematic, restore the timing and magnitude of rigid lever formation, and reduce pathological tissue strain while improving tissue capacity and neuromuscular control.

In pediatric populations, many cases were physiologic and improved with growth and maturation; accordingly, aggressive intervention was often unnecessary in asymptomatic children, and reassurance plus monitoring was commonly appropriate[32]. The clinical focus shifted when symptoms existed (pain, fatigue, reduced activity tolerance) or when function was impaired, particularly during sport and prolonged walking[32].

Foot orthoses were widely prescribed to support the MLA and influence hindfoot and midfoot mechanics. Functionally, orthoses could be considered load-modifying devices that redistribute plantar loads, shifted COP progression, and reduced the magnitude and velocity of pronation by supporting the medial column and controlling rearfoot motion[41]. A medial rearfoot post or wedging strategy could reduce calcaneal eversion and might have reduced internal rotation moments transmitted to the tibia, thereby potentially influencing knee mechanics in individuals whose symptoms relate to

these coupled motions[41]. Critically, orthotic success should have been framed around task-specific outcomes such as improved late-stance stability, reduced pain, reduced fatigue, improved running and landing control, or reduced tissue strain, rather than static arch height alone.

In parallel, strengthening interventions targeting intrinsic foot muscles and neuromuscular control (“foot core”) had gained attention as approaches to enhance active arch support and regulate stiffness during stance (**Figure 5**). This approach was conceptually complementary to orthoses. Orthoses provided immediate external support and load redistribution, whereas strengthening aimed to increase internal capacity to generate stiffness and control pronation timing, which aligned with the “failed rigid lever” model described above[33-35]. Combining orthoses with targeted strengthening might therefore have been particularly relevant for active populations, in whom running and landing magnify the consequences of insufficient arch stiffening and soft-tissue overload.

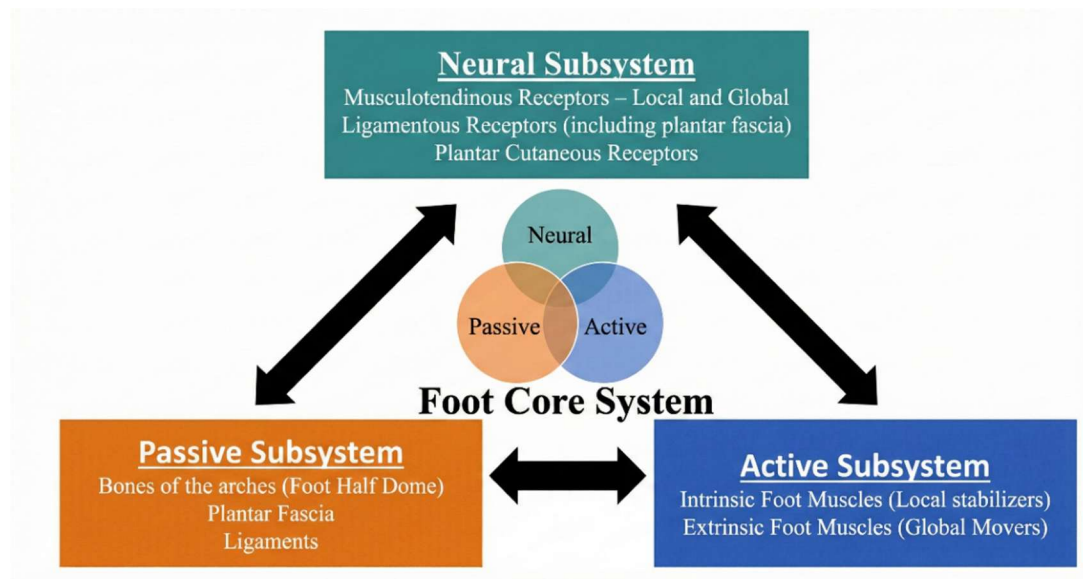


Figure 5 Foot core: interaction of neural, active & passive subsystems[42]

Overall, the goal was not to eliminate pronation, but to optimize motion and loading so that the foot could pronate for shock absorption and resupinate for propulsion at appropriate times[33]. When successful, this reduced strain on the plantar fascia and PTT, improved propulsion efficiency, and might have reduced maladaptive proximal compensations. This biomechanical framing provided a direct rationale for evaluating

orthoses under multiple movement tasks, because different tasks imposed distinct load profiles and might have revealed different mechanisms and benefits of orthotic intervention[33-35,41].

1.2 Mechanism of Action of Foot Orthoses

1.2.1 Shifting Paradigms in Orthotic Therapy

FOs were commonly positioned as the primary mechanical interface between the foot and the ground, and they remained a central non-surgical intervention for symptomatic pes planus. Earlier orthotic practice was strongly influenced by the “Root Theory” , which emphasized placing the subtalar joint (STJ) in a neutral position and seeking an idealized alignment of the kinetic chain. This paradigm assumed that correcting foot “malalignment” would systematically normalize proximal joint mechanics and reduce symptoms. However, modern clinical biomechanics increasingly favored the “Tissue Stress Theory” , which reframed orthotic therapy as a strategy for reducing excessive mechanical stress on specific tissues that were overloaded in a given individual, rather than enforcing a universal skeletal position[43].

Within this framework, the orthosis was prescribed to reduce pathological loading on structures that commonly became symptomatic in flatfoot, including the posterior tibial tendon, plantar fascia, spring ligament complex, and medial midfoot joints. This approach aligned with contemporary evidence that symptom relief often occurred even when changes in gross kinematic alignment were small, suggesting that the primary therapeutic benefit might have come from modifying internal loading conditions rather than achieving visible “correction” . The tissue stress perspective also naturally accommodated inter-individual variability, because two individuals with similar arch morphology might have experienced different symptoms due to differences in tissue capacity, body mass, activity demands, or neuromuscular control.

Mechanistically, FOs could be understood as devices that manipulate the location, magnitude, and timing of ground reaction forces (GRF), thereby modifying external joint moments and the muscular effort required to stabilize the foot and lower limb. By shifting

GRF application and increasing contact area, orthoses could reduce peak stresses on vulnerable tissues while supporting functional movement patterns. In this way, orthoses did not simply “support the arch” , but optimized the interaction between external forces and internal muscle moments in order to reduce fatigue, improve tolerance to repetitive loading, and slow the progression of deformity in at-risk patients[43].

1.2.2 Kinematic Control: The Restriction of Pathological Motion

In symptomatic pes planus, excessive subtalar pronation was typically described as a tri-planar motion that includes calcaneal eversion, talar adduction, and talar plantarflexion. This coupled motion could reduce the congruency of midfoot structures and contributed to prolonged midtarsal joint “unlocking” , which diminishes the foot’s ability to become a rigid lever during propulsion. Orthoses might have provided kinematic control by reducing the magnitude, velocity, or duration of this pronation pattern, especially during the stance phase when tissue loading peaked.

A commonly discussed strategy was the use of a medial heel skive and rearfoot posting to influence calcaneal position and subtalar moments. A medial heel skive modified the heel cup geometry so that contact pressure was increased on the medial plantar heel, which could shift the GRF medially relative to the STJ axis and generate an external supination moment that opposes pronation[44]. This external moment could reduce the tendency for the calcaneus to evert rapidly and could limit end-range valgus positioning, which was often associated with peak tissue strain. Rearfoot posting could achieve a similar goal by altering the interface geometry under the calcaneus, thereby changing the direction and effective lever arm of GRF around the STJ axis[44].

For more severe deformity or when maximal motion control was required, designs incorporated a deep heel cup and high medial and lateral flanges. These features constrained rearfoot and midfoot motion by increasing the rigidity of the orthosis–foot coupling and providing a containment effect around the calcaneus and midfoot. The design could reduce medial talar drift and limit navicular drop by physically resisting

midfoot collapse and rearfoot eversion, which was particularly relevant in pediatric flexible flatfoot and early-stage adult-acquired flatfoot presentations[45].

Kinematic effects at the foot might have propagated proximally through the coupling between subtalar pronation and tibial internal rotation. When orthoses reduced the extent or duration of pronation, they might have reduced tibial internal rotation moments and influence knee alignment dynamics, particularly in individuals who exhibit pronation-related knee valgus patterns. Evidence from 3D motion analysis suggested that even small reductions in peak rearfoot eversion could have been accompanied by clinically meaningful symptom changes, including reductions in patellofemoral pain in some patients whose symptoms are sensitive to changes in tibial rotation and patellar tracking forces[46]. These findings supported the view that modest kinematic changes could still have been therapeutically relevant if they occur at mechanically critical times in stance or reduce end-range motion where tissue strain was highest[46].

1.2.3 Kinetic Control: Pressure Redistribution and Force Modulation

Although orthoses sometimes produced small kinematic changes at the rearfoot, their kinetic effects were often more pronounced and might have been more directly related to symptom relief. In a collapsed flatfoot, plantar loading frequently increased in the medial midfoot and under the first and second metatarsal heads, and COP trajectory might have shifted medially. These loading patterns could elevate local stress on pain-sensitive tissues and contributed to callus formation, joint irritation, and fatigue. Orthoses could address these issues by redistributing plantar pressures, altering COP progression, and modifying the external moments acting around foot joints.

A key kinetic mechanism was the “total contact” concept, in which a custom-molded orthosis increased the effective load-bearing area by conforming to the plantar contour and engaging the midfoot in load support. By increasing the area over which vertical forces were distributed, peak plantar pressures could be reduced according to the pressure relationship $P=F/A$. Pedobarographic studies reported that total-contact insoles reduced peak pressures under the heel and forefoot by transferring part of the load toward the arch

and midfoot regions (**Figure 6**)[47]. This mechanism was particularly relevant in symptomatic flatfoot, where excessive loading of discrete forefoot and rearfoot regions might have contributed to pain and tissue irritation[47].

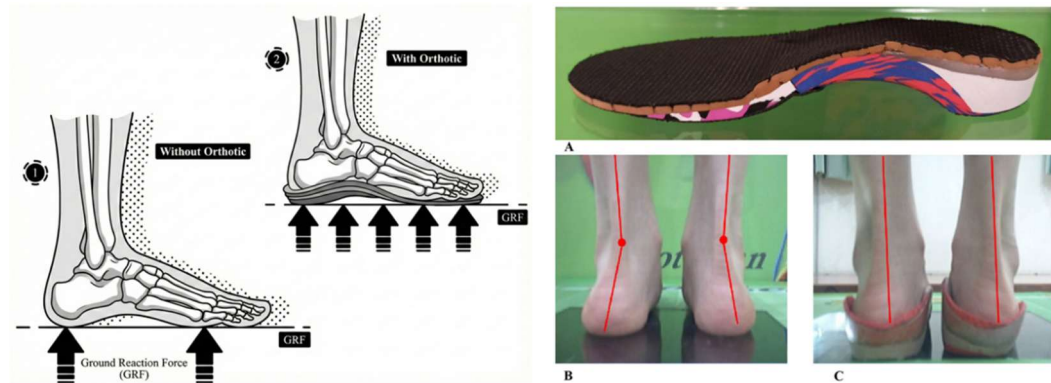


Figure 6 Correcting foot alignment and load distribution with FO[48]

Orthoses might also have produced a lateral transfer of load by supporting the medial column and altering the COP pathway. When the COP was shifted laterally, the lever arm of GRF relative to the STJ axis changed in a way that could reduce the demand on muscles tasked with resisting pronation, particularly the tibialis posterior. This “muscle-sparing” effect could reduce fatigue and eccentric loading demands, potentially lowering the risk of tendinopathy in individuals with limited tissue capacity or high activity volumes[49]. The kinetic viewpoint therefore connected orthotic design to internal muscular effort, suggesting that clinical improvement might have occurred via reduced muscular compensation rather than visible changes in alignment[49].

In addition to force redistribution, material properties and layering contributed to impact attenuation and energy dissipation. Many orthoses incorporated viscoelastic top covers and mid-layers that dampened high-frequency components of impact and reduced localized peak loading rates. Because flatfoot mechanics might have involved prolonged pronation and reduced late-stance stiffening, shock attenuation demands might have increased, and viscoelastic materials could provide partial substitution for compromised natural damping mechanisms. This could reduce the transmission of impact-related forces

up the tibia and might have contributed to symptom relief at both distal and proximal joints, particularly during running and repetitive loading tasks[50].

1.2.4 Neuromuscular and Proprioceptive Mechanisms

Orthotic effects extended beyond structural support and could influence neuromuscular control by altering afferent input from the plantar surface. The plantar skin contained dense mechanoreceptor populations, including receptors sensitive to pressure, vibration, and skin stretch, and these signals contributed to postural regulation and gait control. When an orthosis changed plantar contact patterns, it could modify the sensory information entering the central nervous system (CNS), potentially influencing muscle recruitment strategies and movement patterns.

The concept of a “preferred movement path”, described by Nigg and colleagues, proposed that the neuromuscular system tends to regulate movement in a way that minimizes soft tissue vibration and maintains a stable, economical pattern. From this perspective, orthoses might have functioned as a “sensory filter” that changes plantar input and therefore modifies motor output. For example, increased contact and pressure under the medial arch region may alter timing or magnitude of activation in intrinsic foot muscles or peroneal muscles, contributing to improved active stabilization under load[51]. These effects might have been especially relevant when structural correction was limited by tissue stiffness or high external loads, because neuromuscular modulation could provide an additional pathway for functional improvement[51].

Proprioceptive feedback was also implicated in the clinical observation that even soft or minimally corrective insoles could improve balance and comfort. The tactile presence of an orthosis might have enhanced somatosensory awareness of foot position and loading, encouraging a more stable stance strategy and reducing excessive sway. Experimental studies reported improvements in postural stability measures, including reduced postural sway during single-leg stance, in individuals with flatfoot when sensory input was enhanced by insoles, even when the devices were not designed for strong mechanical control[52]. This suggested that symptom relief and functional gains might

sometimes have arisen from neurosensory pathways rather than purely from changes in skeletal alignment[52].

1.2.5 Advanced Considerations: The Role of AFOs

In more severe or progressive cases, particularly when deformity became rigid or when PTTD had advanced, standard foot orthoses might have been insufficient to control hindfoot valgus and midfoot collapse. When ligamentous and tendinous supports failed, the mechanical leverage required to resist deforming moments might have exceed what an in-shoe device could deliver, especially during higher-demand tasks. In such cases, ankle-foot orthoses (AFOs) could provide additional control by extending above the ankle and creating a longer lever arm to influence the leg-foot complex.

AFOs such as solid or articulated designs, including devices often described in clinical practice (e.g., Richie-type braces), could implement a three-point pressure system that counters valgus collapse and limits excessive hindfoot eversion. By applying stabilizing forces around the ankle and hindfoot, these devices could partially substitute for insufficient posterior tibial tendon function and reduce deformation during stance. This strategy was commonly considered for PTTD and later-stage adult-acquired flatfoot presentations where midfoot collapse was substantial and symptoms persist despited in-shoe orthoses[53].

Material hardness was a critical design variable across both FO and AFO prescriptions. The capacity of an orthosis to resist deformation depended on the hardness of its shell and the geometry of its arch support, which were influenced by material selection and thickness distribution. Heavier patients or individuals with more rigid deformities often required stiffer materials (e.g., higher-density polymers or carbon composites) to generate sufficient supination moments and resist collapse under load. Conversely, individuals with flexible flatfoot and lower body mass might have benefited from semi-rigid designs that provide support without excessive discomfort and that allow some natural motion and shock absorption[54]. Modern CAD or CAM and additive manufacturing approaches had enabled variable hardness designs, in which thickness and

material properties could be spatially tuned to increase support in the medial arch while maintaining comfort and compliance in regions such as the heel[54].

1.2.6 The “Kinematic Paradox” and Clinical Outcomes

Orthotic research repeatedly observed that patients could report meaningful reductions in pain and improved functional capacity despite relatively small measured changes in foot and ankle kinematics. This phenomenon was termed the “kinematic paradox” , reflecting the apparent disconnect between limited changes in skeletal alignment and substantial clinical benefit[55]. The paradox suggested that clinical outcomes might have been more sensitive to reductions in tissue stress, changes in internal joint moments, and altered load distribution than to visible changes in rearfoot angle or arch height during dynamic tasks.

A plausible explanation was that orthoses might have reduced internal stress by preventing the foot from reaching end-range pronation positions where strain on tissues such as the plantar fascia and posterior tibial tendon was highest. Even if peak calcaneal eversion changed by only 1–2 degrees, the orthosis could still have reduced the rate of eversion, shorten the duration of excessive pronation, or alter the timing of loading in a way that meaningfully reduces cumulative tissue load. Orthoses might also have reduced compensatory muscular demand by changing GRF lever arms, which could reduce fatigue and improve endurance even when kinematic metrics appeared largely unchanged[55,56].

This interpretation aligned with the tissue stress paradigm, in which success was evaluated by symptom reduction, improved tolerance to activity, and functional performance rather than by radiographic or kinematic “correction” alone[43,56]. Consequently, clinical assessment should have emphasized patient-reported outcomes, task-specific function, and measures of loading where possible, because orthoses might have operated primarily by shifting forces into a “zone of safety” that allows tissue recovery and reduces recurrent overload[56]. In this sense, the kinematic paradox was not a contradiction but a reminder that biomechanics was fundamentally about the interaction

of motion and force, and that therapeutic benefit could have arise from substantial kinetic changes occurring alongside modest observable kinematic shifts[55,56].

1.3 Biomechanical effects in diverse movement tasks

1.3.1 Walking

1.3.1.1 Joint kinematics

Study found that arch support insoles led to smaller peak everted position and reduced rearfoot motion compared to normal insoles[57]. Another study reported that orthotic insoles improved the arch index both before and after long-term intervention during walking on level surfaces and stairs[58]. In comparison to wearing shoes only, using 3D printed foot orthoses resulted in increased peak ankle dorsiflexion[59]. Additionally, rigid 3D printed foot orthoses reduced midfoot eversion and forefoot abduction, while 3D printed foot orthoses with posting decreased ankle eversion angle[60]. In terms of arch support, 3D printed orthoses showed a smaller arch height drop compared to traditional orthoses[61]. Furthermore, insoles were found to increase navicular height, and 3D printed insoles specifically increased ankle dorsiflexion angle, with the wedge condition being most effective[62]. The maximum ankle dorsiflexion angle during mid-stance, the maximum ankle external and internal rotation angles, and the maximum ankle plantarflexion angle during loading response were all reduced under orthotic settings when compared to wearing shoes alone. Furthermore, when compared to the 15° inverted angle and shoes-alone situations, the maximal ankle plantarflexion angle at toe-off only reduced in the 25° inverted angle condition. In contrast, under orthotic settings, the maximal hip external rotation angle rose[63]. It is worth noting that one study by Karimi did not discover any appreciable variations in lower limb kinematics[64].

The vast majority of studies had found that foot orthoses have an impact on walking kinematics. During walking, foot orthoses provided better arch support compared to no insoles and have a significant effect when walking up and down stairs[63,65]. Foot orthoses reduced peak ankle eversion angle, maximum ankle external rotation angle, internal rotation angle, and increased maximum hip external rotation angle[57,66]. These

kinematic changes, which help to reduce the abnormal posture of flatfoot, help flatfoot patients to perform walking. However, the effects produced by different designs of foot orthoses on walking kinematics were controversial. Some studies had found that the use of 3D printed foot orthoses can reduce the arch height drop better compared to traditional foot orthoses, but the effect of different designs of 3D printed insoles on the reduction of the arch height drop was not consistent. The wedge insoles had the best effect, and the effect of the other designs of 3D printed insoles had no difference[50,65]. The effects of different designs of 3D printed insoles on reducing peak ankle eversion angle were also inconsistent. The 3D printed foot orthoses with posting was the most effective, but there was no difference in effect between the other designs of 3D printed insoles. The rigid 3D printed foot orthoses, compared to no insoles and other types of 3D printed insoles reduced midfoot eversion and forefoot abduction[67]. These suggested that changing the design of the insole may increase its alteration of walking kinematics and better correction of abnormal foot posture.

However, there were also opposing trends in the effects of different designs of foot orthoses on walking kinematics. Foot orthoses reduced the range of motion in the sagittal plane of the rearfoot, maximum ankle plantarflexion and dorsiflexion angle, while angle foot orthoses with a 25° inverted further reduced the maximum ankle plantarflexion angle[57,66]. However, 3D printed insoles increased peak ankle dorsiflexion angle and the effect of different designs of 3D printed insoles on increasing the peak ankle dorsiflexion angle was not consistent, with wedge insole increasing the most significantly but there was no disparity in the effect between other designs of 3D printed insoles[65]. There was still controversy about how ankle angles change to benefit population with flatfoot. One view was that flatfoot was due to excessive dorsiflexion and eversion resulting in talus and tarsal subluxation and pronation, where dorsiflexion or hypermobility of the ankle should be controlled[68]. Another view was that the limited range of motion of the ankle leads to rearfoot inversion and that the range of motion of the ankle should be restored[69]. Therefore, it remained to be explored which design of foot orthoses was more effective in modifying ankle dorsiflexion and extension.

1.3.1.2 Joint kinetics

Studies examined how insole interventions impact lower limb joint kinetics during walking[57]. Arch support insoles resulted in lower peak evtor moments compared to normal insoles[64]. Foot insoles decreased the first and second peaks of adductor moments on the ankle joint. 3D printed FOs reduced the maximum ankle evtor moment by an average of 35% and significantly decreased the peak external rotator moment by 16%. Additionally, the maximum ankle plantar flexor moment with 3D printed FOs was significantly higher than with regular shoes[70]. FOs with posting reduced ankle inversion moment and increased knee abduction moment[60]. Traditional and 3D printed foot orthoses showed lower ankle plantar flexion moment during control conditions and higher power absorption compared to control and traditional conditions[61]. The study of Hsu was the only one that did not find differences in lower limb kinetic[62].

The vast proportion of studies had found that foot orthoses influenced walking joint kinetics. Foot orthoses exhibited smaller peak ankle eversion moment and ankle power absorption; 3D printed insoles similarly exhibited reduced maximal ankle eversion moment, peak external rotator moment, and ankle power absorption[57,61,71]. The magnitude of joint moments during walking can be considered key to preventing injury, and using fewer moments will reduce the likelihood of injury from overuse[72]. Increasing the stiffness of 3D printed Fos will reduce the ankle inversion moment and the functional demands on the inversion muscles will be reduced, which may result in overuse injuries to other compensatory muscles. However, a rise in the knee abduction moment, despite values within the safe range, may have an undesirable effect and lead to the development of osteoarthritis in the medial compartment knee[67]. Like joint kinematics, opposing results existed in studies regarding ankle sagittal plane moment. 3D printed insole showed greater ankle plantarflexion moment, whereas traditionally made insoles and 3D printed insoles showed smaller ankle plantarflexion moment[61,71]. The divergence of the studies lies in the understanding of muscle activation of the peroneus longus, peroneus brevis, soleus, medial peroneus, and lateral peroneus in patients with flatfoot during the stance phase, which can be further validated by electromyogram in subsequent studies.

1.3.1.3 Plantar pressure and kinetics

Articles reported the impact of insole interventions on the planter pressure during walking. The long-term intervention study of insoles showed following. Peak force in the first metatarsal region and peak pressure in the third metatarsal region were initially lower in the ethylene-vinyl acetate (EVA) insoles group than in the standard insoles group. By week 8, though, the group using the EVA insoles had more midfoot force and peak pressure. Conversely, by week 8, the first metatarsal area's peak force was greater and the fourth and fifth metatarsal areas' peak pressures and forces were lower in the traditional insoles group than they had been in the previous week[73]. Regarding flatfoot, orthotic insole treatment effectively decreased plantar pressure, especially during downhill walking[58]. In arch support insoles, the peak pressure of the big toe (BT) was higher on uphill and level surfaces compared to flat insoles, while the peak pressure of the mid-heel (MH) was lower on all surfaces. Additionally, the peak pressure of the second (M2), third (M3), and fourth (M4) metatarsals was higher in arch-support insoles than flat insoles, while the lateral heel (LH) pressure was lower in arch-support insoles[6]. In comparison to the control, there was a rise in peak pressure at the medial midfoot region and a fall in the rearfoot region for the insole conditions.

At modest walking speeds, the peak pressure of plantar pressure redistribution insoles (PPRI) in the toe (T) region was higher than the flat insole. Peak pressure in the M2 region was lower when walking slowly with the PPRI insole than when using the orthotic insole. Peak pressure in the MH region was lower when walking at slow, normal, and rapid walking speeds with the PPRI insole than when walking at slow walking speeds with the orthotic insole. Furthermore, at a normal walking speed, the peak pressure in the LH area was lower while using the PPRI footbed than when using the flat insole. At normal, fast, and slow walking speeds, the peak pressure with the orthotic insole was lower than with the flat insole[74]. The mean pressure and force at the second, third, fourth, and fifth metatarsophalangeal (MTP2,3,4,5) joints as well as the heel decreased with the usage of insoles. Nevertheless, it was discovered that wearing insoles increased pressure in the medial midfoot. There were no discernible variations in plantar pressure and force

between the rigid insole that was custom-molded and the prefabricated soft insole[71]. Walking with foot orthoses caused the greatest alterations in the midfoot area, where peak pressures, mean pressures, and contact area all increased[75]. In terms of pressure on the middle foot, both INS2 (customized thermoplastic) and INS3 (3D printed insole) result in increased peak pressure when compared to INS1 (original insole), with INS2 having a 35% higher peak pressure and INS3 showing a 23% higher peak pressure in the middle foot region. For the rearfoot pressure, INS2 decreased the peak pressure when compared to the INS1[66].

Foot orthoses affected plantar pressure and kinetics during walking. The impact of foot orthoses on plantar pressure and kinetics typically included a rise in peak pressure at the midfoot and a reduction in peak pressure at the heel[73,74]. Foot orthoses helped maintain the calcaneus in a normalized alignment, improving pressure distribution and absorption. With the addition of longitudinal arch support, these insoles shifted pressure from the rearfoot to the midfoot and decreased rearfoot pressure, ultimately reducing the risk of tibial stress fractures[76]. The intervention of foot orthoses can normalize plantar pressure distribution. Different designs of foot orthoses caused further effects on plantar pressure distribution. Customized insoles and 3D printed insoles resulted in a further increase in midfoot peak pressure, a decrease in rearfoot peak pressure, and a further increase in midfoot peak pressure after long term intervention[62,75,77]. Peak pressure in the metatarsal region is still a controversial topic. According to recent research, using a medial longitudinal arch support may help to transfer pressure from the first metatarsal to the second and third metatarsals in the central forefoot. Additionally, this support may lessen forefoot pronation and lessen strain on the first metatarsal by keeping the subtalar joint in a position that is similar to its usual state[78]. After long term intervention, 3D printed insoles reduced peak pressure and force in the medial metatarsal region[62]. However, it had been shown that foot orthoses had no impact on pressure and force distribution in the M1 region[73,74,77]. Subsequent studies should pay more attention to this point. Different walking styles and different walking speeds were also factors that

influenced the alteration of plantar pressure by foot orthoses. Pressure distribution in the rearfoot and metatarsal region was altered as a result[63,64,70].

1.3.1.4 Ground reaction force

Articles investigated the effects of insole interventions on ground reaction forces during walking. One study discovered that using arch support insoles while walking led to a reduction in the second peak of vertical force, but an increase in the first peak of anteroposterior force. In addition, the sideways component of the force exerted by the ground decreased when participants utilized insoles, as opposed to walking without them[64]. Another study reported lower vertical ground reaction forces during walking with both Traditional and 3D printed insoles compared to a control condition[61]. The studies of Han, Lin and Cheng did not find differences in ground reaction force[57,59,70].

There were not many studies related to the effect of foot orthoses on ground reaction forces during walking and most of them reported no difference existed[57,61]. Studies showed that aligning the foot structure with foot orthoses resulted in a decrease in the peak vertical GRF[59,71]. Further studies should follow to determine the effects.

1.3.2 Running

Articles investigated the effects of insole interventions on lower limb joint kinematics during running. In the first study, the use of heel lifts (HL) caused the peak dorsiflexion angle to drop in comparison to the control condition, whereas the use of custom arch support orthotics (CASO) increased it. Furthermore, when employing CASO, the peak dorsiflexion angle was greater than when using HL[67]. In the second study, when compared to wearing standard shoes (SHOD), FOs decreased the dorsiflexion angle and decreased ankle eversion during the whole stance phase. During the stance phase, FOs increased the knee adduction angle at the knee joint. Furthermore, in comparison to the Low-Dye Taping (LDT) condition, FOs enhanced hip external rotation in the horizontal plane at the start of the stance phase[65].

Articles reported the impact of insole interventions on the kinetic of lower limb joints during running. The peak plantarflexion moment was reduced when running with both CASO and HL compared to the control condition. Additionally, peak achilles tendon loading (ATL) was lower when using HL during running compared to the control. In comparison to the control, both the CASO and HL interventions demonstrated a marginally longer duration from initial contact to peak Achilles tendon force (duration to peak Achilles tendon force). Running with HL resulted in a lower achilles tendon loading rate (ATLR) compared to the control group[67]. The plantar flexor moment was lower with FOs compared to SHOD. Knee moments (extensor, external rotation, and abductor) were all higher with Fos. Additionally, the knee external rotation moment remained higher with FOs when compared to LDT. In contrast, the hip extensor moment was higher with LDT than with FOs and SHOD[65].

Foot orthoses had been shown to correct abnormal foot plantarflexion moment, decrease achillea tendon forces, and redress abnormal ankle eversion foot posture during running, but adding the medial heel wedge results in increased knee moment. Like walking, what effect foot orthoses have on the peak dorsiflexion angle still need to be further investigated, and different designs of foot orthoses affected the peak dorsiflexion angle differently[6,60].

1.3.3 Jumping and Landing

The impact of insole interventions on lower limb joint kinematics during jumping was examined in previous research. Study found that foot orthoses resulted in less ankle eversion during the take-off phase of a countermovement jump (CMJ) compared to an insole condition. Another study reported that during a standing broad jump (SBJ), foot orthoses were associated with less ankle eversion compared to the insole condition[77]. Modified wedge foot orthoses (MWFOs) were found to reduce the midfoot dorsiflexion angle during the landing phase of drop landings on a level surface when compared to shoe wear. When using MWFOs as opposed to shoes, a reduced midfoot abduction angle was also observed. Furthermore, MWFOs and total contact foot orthoses (TFOs) both led to

a smaller ankle eversion angle compared to shoes, with MWFOs also showing a smaller ankle eversion angle compared to TFOs. During drop landings on a valgus-inclined surface, both MWFOs and TFOs were associated with a smaller midfoot dorsiflexion angle compared to wearing shoes. Additionally, MWFOs showed a smaller midfoot dorsiflexion angle compared to TFOs. Moreover, MWFOs resulted in a smaller midfoot abduction angle compared to TFOs and shoes. Both MWFOs and TFOs were associated with a smaller ankle eversion angle compared to shoes, with MWFOs also showing a smaller ankle eversion angle compared to TFOs. Compared to TFOs, MWFOs showed a larger hip flexion angle, and compared to TFOs and shoes, MWFOs showed a smaller hip internal rotation angle[79].

Articles reported the impact of insole interventions on the kinetic of lower limb joints during jumping. Foot orthosis conditions was less peak ankle frontal moment than insole condition at SBJ[77]. Dropping onto a level surface: A smaller ankle inversion moment was observed for individuals MWFOs compared to those wearing shoes (SHOD). A smaller hip abduction moment was observed for individuals wearing toe-only footwear (TFOs) compared to MWFOs. Dropping onto a valgus inclined surface: A smaller hip abduction moment was observed for both TFOs and MWFOs compared to those wearing shoes[79]. Research reported the impact of insole interventions on GRF during jumping. Foot orthosis conditions was lower peak horizontal GRF than insole condition at SBJ[77].

In jumping and landing motions, foot orthoses can reduce foot and ankle eversion and increase foot stability. In addition, foot orthoses reduced ankle stress and enhance hip stability in landing motion. Different designs of foot orthoses have different effects on hip joint biomechanics[58,79].

Overall, recent studies have demonstrated the effectiveness of foot orthoses during walking, running, jumping and landing motions, correcting abnormal foot postures, and better protecting patients for these motions. However, there was still disagreement about changes in peak ankle eversion angle, ankle sagittal plane moment, and peak pressure in the metatarsal region. No consensus on whether the biomechanical effectiveness of the

insole can be further enhanced using personalization, 3D printed, addition of a variety of accessories, and adjusting material parameters, which could be further explored in subsequent studies to find the most effective foot orthoses. In addition, follow-up studies could focus more on the comparison between FO and other interventions, analyzing their advantages and limitations during gait cycle, and assessing FO intervention stability and regulatory mechanisms under high-impact loads like landing from different heights.

1.4 Digital Design Strategies for Foot Orthoses

1.4.1 Paradigm Shift: From Artisan Craft to Digital Engineering

The fabrication of foot orthoses is increasingly transitioning from a manual, technician-driven craft to a standardized digital engineering workflow that is based on measurable geometry, traceable design rules, and repeatable manufacturing. Traditional fabrication methods commonly involve plaster casting, positive model rectification, and thermoforming, and each step introduces opportunities for dimensional distortion and operator-dependent variation. Plaster casting can compress soft tissue and is sensitive to the clinician's handling technique, while positive mold modification depends heavily on tacit experience and is often difficult to document precisely. This creates a reproducibility challenge because two practitioners can produce materially different orthotic shapes for the same foot, which can plausibly influence pressure redistribution and perceived comfort in clinical use[80].

Digital workflows that sit within the CAD or CAM ecosystem address these limitations by formalizing the orthosis as a digital object that can be stored, audited, shared, and reproduced. Digitization enables clinicians to preserve baseline designs, implement incremental modifications across visits, and quantify design changes such as arch height, posting angle, or heel cup depth as explicit parameters rather than implicit "hand skills". In addition, digital design reduces material waste because it eliminates plaster and reduces iterative remakes caused by casting or rectification errors. Digital manufacturing also expands the design space by allowing complex geometries, including variable-stiffness structures, that are difficult or impossible to create reliably through

subtractive milling or layered foam lamination. This paradigm shift therefore changes orthoses from handcrafted artifacts into engineered devices that can be systematically optimized, compared, and scaled across patient populations while still supporting customization at the individual level (**Figure 7**)[80].

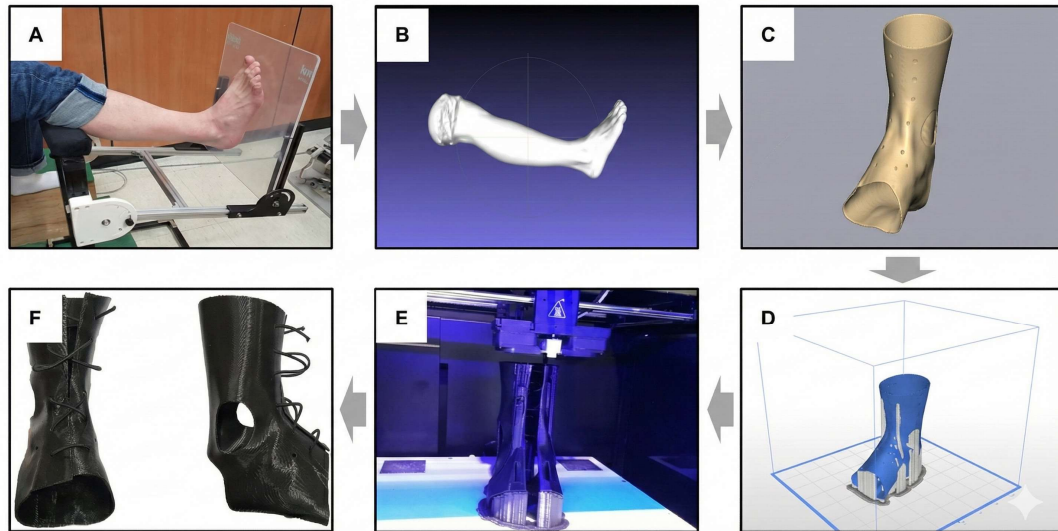


Figure 7 From 3D scanning to final 3d-afo prototype[81]

1.4.2 Data Acquisition: Precision Through 3D Scanning

Accurate data acquisition is the foundation of digital orthotic design because the geometry of the plantar surface and the spatial relationship between the rearfoot and midfoot determine how well a device achieves total contact and how reliably it transfers load away from vulnerable tissues. Contemporary clinical scanning most commonly uses optical methods such as structured light and laser line scanning, both of which reconstruct 3D surface topology through triangulation. Structured light systems project known patterns onto the foot and infer depth from pattern deformation, while laser systems sweep a line and reconstruct surface points from captured reflections, and both approaches reduce the mechanical distortion that can occur when plaster physically contacts and compresses tissue (**Figure 8**)[82].

Reliability depends not only on scanner resolution but also on scanning protocol, including patient positioning, weight-bearing condition, and stabilization of subtalar

posture during capture. Studies comparing 3D scanning with plaster casting suggest that when a positioning device is used and the subtalar joint is controlled, digital scans can reproduce foot geometry with small deviations, supporting their use for high-fidelity custom devices where millimeter-level differences may affect fit and pressure distribution [83]. This level of geometric precision is especially important for total-contact designs intended to redistribute pressure in neuropathic or high-risk feet because small local mismatches can create focal pressure points rather than offload them.

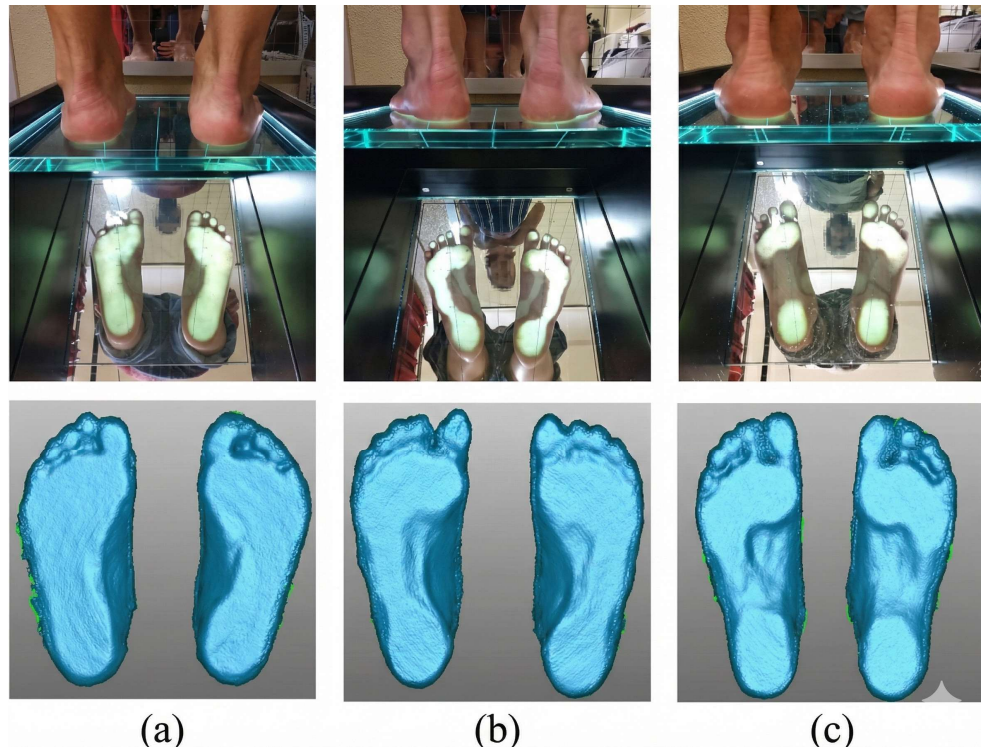


Figure 8 3D Scans across arch types[84]

A major advantage of digital workflows is the ability to integrate static geometry with dynamic functional data. Pedobarography can identify task-specific high-pressure regions and COP progression, and these data can be spatially mapped onto the 3D foot model so that offloading features are placed where load actually occurs during gait rather than where clinicians infer load from static posture alone. This fusion of 3D shape with dynamic pressure signatures supports more targeted modifications such as metatarsal bars, local depressions, or medial arch relief, and it enables iterative refinement that is anchored to measurable biomechanics rather than trial-and-error adjustments[85].

1.4.3 Computational Design: Parametric Modeling and Generative Algorithms

Once a 3D scan is obtained, the design phase converts the anatomical model into an orthosis that encodes both fit and function. Early CAD workflows often resembled “virtual carving”, in which a technician manually shaped surfaces to approximate traditional plaster rectification. Modern workflows increasingly use parametric modeling, where the orthosis is controlled by explicit variables such as arch height, medial flange height, heel cup depth, shell thickness, and posting geometry. Parametric design supports rapid, traceable iteration because a clinician can adjust a single parameter and regenerate the entire model while preserving alignment to the scan-derived plantar surface. This approach also reduces dependence on individual operator style and promotes reproducibility across clinics and manufacturing sites, which is a core goal of digital engineering strategies in orthotics[86].

Generative algorithms extend parametric design by searching a broad space of possible geometries to meet explicit objectives, such as minimizing mass while maintaining stiffness, reducing predicted peak plantar pressure in a target region, or maximizing durability under cyclic loading. Topology optimization is one such approach, in which the algorithm determines where material is structurally necessary and where material can be removed without compromising performance. The resulting structures often appear organic because they reflect efficient load paths rather than traditional uniform shells, and they can produce improved strength-to-weight ratios that are especially beneficial for performance-focused devices[87].

FEA provides the computational link between geometry and mechanical behavior because it allows designers to simulate stress, strain, and deformation under boundary conditions that approximate stance or gait loading. In orthotic design, FEA can be used to compare posting angles, arch stiffness, or relief regions before manufacturing, thereby reducing the number of physical prototypes and clinical trial fittings required. Work that uses FEA to evaluate design variants demonstrates how computational simulation can be used to predict whether a device is likely to reduce strain in key structures such as the

plantar fascia, and it supports a more rational, hypothesis-driven design workflow rather than relying solely on empirical adjustments after discomfort occurs (**Figure 9**)[88].

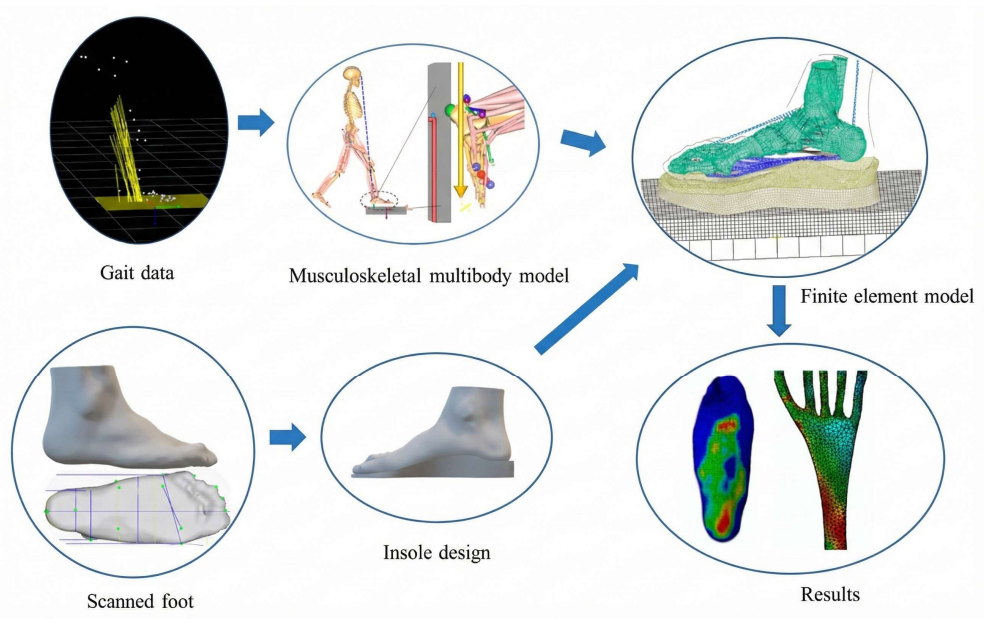


Figure 9 Overview of the foot-ankle complex and insole finite element model[12]

1.4.4 Additive Manufacturing: Complexity Without Cost

Additive manufacturing (AM) enables a “complexity is free” principle because the machine cost is driven more by build volume and time than by geometric intricacy. This matters for orthoses because biomechanical needs are rarely uniform across the foot, and the ideal device often requires different stiffness profiles across the rearfoot, midfoot, and forefoot. AM supports this by allowing internal lattice structures and spatially variable infill patterns, which can tune local compliance and energy return without introducing bonding interfaces that can fail in laminated constructions. Designers can adjust lattice unit-cell geometry, wall thickness, and infill density to create region-specific stiffness gradients, allowing a single-piece orthosis to provide stable rearfoot control, supportive midfoot stiffness, and forefoot compliance for comfort and toe-off mechanics (**Figure 10**)[89].

Material selection and printing method strongly determine performance, durability, and long-term creep behavior. Fused filament fabrication (FFF) is accessible and cost-effective and can print polymers such as thermoplastic urethane (TPU) that provide

flexibility and fatigue resistance, which can be attractive for devices targeting comfort and moderate control in flexible flatfoot populations[90]. Selective laser sintering (SLS) can produce nylon parts with more uniform mechanical properties and high durability, and it is often considered suitable for functional orthoses that must resist repeated loading while maintaining geometry across time. SLS also supports batch production, which reduces per-unit cost when throughput is high and enables scalable customization workflows in clinical practice[91].

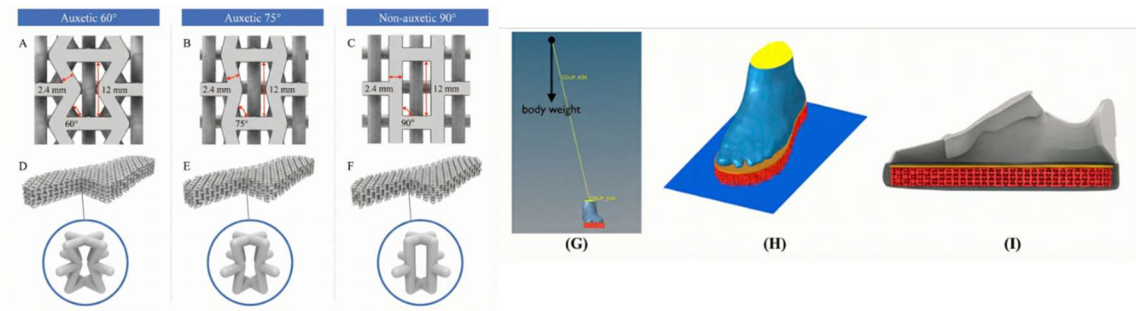


Figure 10 Lattice midsole: from unit cell design to FEM analysis[92]

Clinical and mechanical comparisons between 3D-printed devices and conventional polypropylene orthoses suggest that printed orthoses can achieve comparable functional control while providing additional benefits related to weight reduction and ventilation through perforated or lattice-based designs. Studies reporting improvements in comfort ratings alongside measurable effects on rearfoot motion provide evidence that AM does not merely replicate traditional devices but can offer distinct user-experience advantages that may support adherence in long-term use[73]. At the same time, AM introduces new engineering considerations, including printer calibration, anisotropy in certain printing processes, surface finishing, and quality assurance protocols, all of which must be managed to ensure that the delivered device matches the intended mechanical behavior and fits the patient reliably[73,89-91].

Overall, the transition from artisan craft to digital engineering has established a robust foundation for modern orthotic fabrication, characterized by the geometric precision of 3D scanning, the iterability of parametric modeling, and the structural complexity enabled by additive manufacturing. Digital workflows can effectively

quantify design parameters and utilize FEA to predict mechanical behavior. Research can integrate machine learning algorithms to establish an efficient, data-driven engineering approach for highly personalized FO parametric design.

2 MATERIAL AND METHODS

2.1 Overview of the experimental and computational workflow

This work integrated biomechanical experimental testing with musculoskeletal modeling, Bayesian statistical analysis, vector coding techniques, continuous wavelet transforms, finite element simulations, and machine learning algorithms. It elucidated the biomechanical effects of FOs for flatfoot across diverse movement tasks (walking, running, and landing) and established a personalized, integrated design workflow for FOs.

Figure 11 showed the overview of our works.

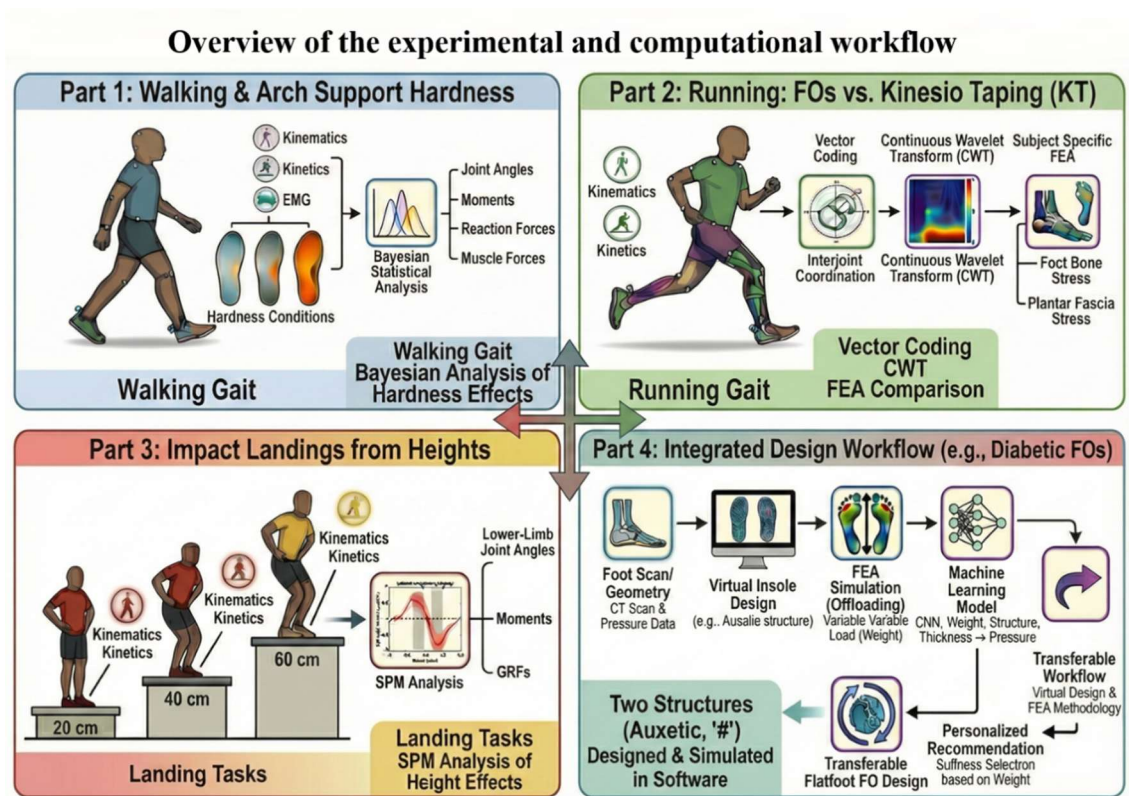


Figure 11 Experimental and computational workflow

Part 1 investigated the hardness of FOs arch support. All participants completed walking gait trials in a standard biomechanics laboratory under different hardness

conditions. Kinematic and kinetic data were collected, and electromyography (EMG) was recorded synchronously for subsequent musculoskeletal modeling and parameter estimation. A Bayesian approach was used to evaluate the effects of arch-support hardness on lower-limb joint angles, joint moments, joint reaction forces, and muscle forces.

Part 2 comparatively analyzed the biomechanical benefits of FOs and KT. Participants performed running gait trials under different intervention conditions, and kinematic and kinetic data were collected. Musculoskeletal modeling was used to compute knee and ankle joint angles and moments. An optimization-based vector coding method quantified inter-joint coordination and variability between the knee and ankle. Continuous wavelet transforms (CWT) were applied to characterize the time–frequency features of multi-directional GRF during the stance phase, followed by Bayesian statistical analysis. For computational simulations, reverse engineering was used to obtain the right-foot geometry of one participant and to build subject-specific foot–ankle finite element models, which quantified stresses in foot bones and the plantar fascia under different intervention conditions.

Part 3 examined how the biomechanical benefits of FOs changed under impact landings from different heights (20, 40, and 60 cm). Kinematic and kinetic data were collected and calculated across landing conditions, and assessed the effects of landing height on stance-phase.

Part 4 developed an integrated FO design workflow using diabetic foot orthoses as an example. FEA simulated the offloading effects of 3D-printed insoles with different structural designs on plantar pressure in patients with diabetic foot. Machine learning models were then trained to recommend effective personalized insole solutions for patients with different weights. The resulting workflow was transferable and could inform the design of flatfoot FOs.

Based on these efforts, this work helped optimize flatfoot FO design, revealed the advantages and limitations of FO interventions, clarified the effectiveness of FOs under

high-impact intensities. The developed workflow could inform flatfoot FO design while providing an efficient framework for future orthosis development.

2.2 Ethics statement

This dissertation was performed in compliance with the declaration of Helsinki and was approved by the Institutional Ethics Committee of Ningbo University. Before taking part in this study, all human subjects and the animal owners were informed of the corresponding experiment and simulation content as well as the potential risks and then they gave the consent to participate, as shown in **Figure 12**.

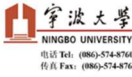
 Research Academy of Grand Health, Ningbo University Human Informed Consent Form Instructions to the Student Researcher(s): An informed consent/assent/permission form should be developed in consultation with the Adult Sponsor, Designated Supervisor or Qualified Scientist. This form is used to provide information to the research participant (or parent/guardian) and to document written informed consent, minor assent, and/or parental permission. - When written documentation is required, the researcher keeps the original, signed form. - Students may use this sample form or may copy ALL elements of it into a new document. If the form is serving to document parental permission, a copy of any survey or questionnaire must be attached. Student Researcher(s): _____ Title of Project: _____ I am asking for your voluntary participation in my science fair project. Please read the following information about the project. If you would like to participate, please sign in the appropriate area below. Purpose of the project: _____ If you participate, you will be asked to: Time required for participation: _____ Potential Risks of Study: _____ Benefits: _____ How confidentiality will be maintained: _____ If you have any questions about this study, feel free to contact: Adult Sponsor/OS/DS: _____ Phone/email: _____ Voluntary Participation: Participation in this study is completely voluntary. If you decide not to participate there will not be any negative consequences. Please be aware that if you decide to participate, you may stop participating at any time and you may decide not to answer any specific question. By signing this form I am attesting that I have read and understand the information above and I freely give my consent/assent to participate or permission for my child to participate. Adult Informed Consent or Minor Assent Date Reviewed & Signed: _____ Research Participant Printed Name: _____ Signature: _____ Parental/Guardian Permission (if applicable) Date Reviewed & Signed: _____ Parent/Guardian Printed Name: _____ Signature: _____	 宁波大学 NINGBO UNIVERSITY 中国 浙江 宁波市江北区风华路818号 邮编: 315211 No.818, Fenghua Road, Ningbo, Zhejiang, China 315211 电话 Tel: (086)-574-8760253 传真 Fax: (086)-574-8760461 Research Academy of Grand Health Ningbo University Institutional Review Board Procedures At the Research Academy of Grand Health, Ningbo University, all human subjects' research activities come under the purview and oversight of the Research Support Services Office and the Institutional Review Board (IRB), irrespective of whether the research is funded or non-funded, minimal risk or more. The Human Subjects Protection policy applies to all Research Academy of Grand Health, Ningbo University faculty, staff, and students conducting human subjects' research on or off-campus (domestic or international sites) as well as visitors conducting research at the Research Academy of Grand Health Ningbo University. Researchers, including the faculty advisors of student researchers, must successfully complete online human subjects' protection training before submission to the IRB. Human subjects protection is a collaborative effort by the researcher and the Research Academy of Grand Health, Ningbo University. The IRB is charged with the responsibility of protecting the rights and welfare of human subjects involved in research. The composition of the IRB and the number of members on the committee are in accordance with national regulations. IRB members are appointed by the Vice Provost for Research on the recommendation of the chairperson of the IRB. Members are appointed for renewable, three-year terms and include faculty and community members with expertise in the various disciplines engaged in human subjects' research on campus. Scope of Review The IRB reviews research involving human subjects if one or more of the following apply: - the research is sponsored by CN, regardless of the location of the project; - the research is conducted by, or under the direction of, any staff, faculty, student, or other agent of CN in connection with his or her institutional responsibilities; - the research is conducted by or under the direction of any employee or agent of CN using any property or facility of CN; - the research involves the use of CN's non-public information to identify or contact human research subjects or prospective subjects.
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Figure 12 The human informed consent form and institutional review board procedures

2.3 Participants

In part 1, participants were primarily recruited through posters on campus, emails, social media announcements, personal invitations. Inclusion criteria were as follows: (1) females aged 18–40 years with flexible flatfoot, defined as the navicular drop greater than 10 mm and the arch height index less than 0.31, with the arch height index calculated by dividing the dorsal height measured at the foot's midpoint by the truncated foot length

from the heel to the first metatarsophalangeal (MTP) joint, and all measurements taken in a standardized standing position with equal weight distribution[93], (2) right-leg dominance, (3) no lower-extremity injury in the previous six months, (4) no history of lower-extremity or spinal surgery, (5) no musculoskeletal or neurological disorders that could affect movement, (6) all participants were asymptomatic and had never used FO, in order to avoid potential influences of symptoms or prior FO use on biomechanical outcomes. 20 participants who met the inclusion criteria were ultimately enrolled with the following characteristics: body mass 52.34 ± 3.51 kg, height 162.14 ± 3.76 cm, age 21.92 ± 2.32 years, and foot length 24.32 ± 1.19 cm. Sample size was chosen to match typical ranges in biomechanics Bayesian research[94-96]. Current research was approved by university's institutional review board, and participants provided informed consent. In part 2 and 3, participants were recruited through campus advertisements and social media. The inclusion criteria were the same as those used in Part 1. A total of 24 participants were recruited (12 males and 12 females). For the male group, the mean $\pm$ SD age was 23.4 ± 2.1 years, height was 177.2 ± 5.8 cm, weight was 72.5 ± 7.4 kg, and foot length was 26.5 ± 1.1 cm. For the female group, the mean $\pm$ SD age was 22.8 ± 2.3 years, height was 164.3 ± 5.4 cm, weight was 57.6 ± 6.1 kg, and foot length was 24.1 ± 0.9 cm. In part 4, a healthy male volunteer (shoe size = European 42, age = 26 years, height = 174 cm, weight = 78 kg) voluntarily contributed lower limb imaging data for simulating the foot modeling of asymptomatic diabetic foot patients.

2.4 Foot orthosis

In part 1, we customized FO for each participant based on individual foot characteristics, as illustrated in **Figure 13A**. The FO were fabricated individually to correct pathological alignment and provide functional support. With the foot kept non-weight-bearing, an experienced clinician first manually positioned the foot into a standardized neutral corrected posture (e.g., subtalar neutral) and maintained the medial longitudinal arch at a clinically appropriate height based on anatomical landmarks. The corrected foot posture was then captured using a 3D scanner (easy-foot-scan, Ortho Baltic) and used as the template for FO fabrication. The FO design, including overall shape and

arch support, was created using Blender (Version 4.3.2, Blender Foundation, Netherlands). The EVA base (Shore A 50) and TPU arch support (Shore C 30, 50, 70) were 3D-printed (Dragon(L) 3D Printer, WINBO). Several research had used medium-hardness EVA of approximately Shore A 50 as the base in order to balance cushioning and support[97,98]. With regard to the choice of hardness levels, we compared relatively soft and hard values reported in previous research and commonly used hardness levels, and also drew on the experience of orthotists[99,100]. Additional design parameters were as follows: the medial longitudinal arch height was 20.80 ± 2.98 mm, and the heel cup depth was 15.21 ± 2.19 mm. The first main loading region was located at the medial longitudinal arch, beneath the navicular and medial cuneiform bones, where the arch segment was reinforced and elevated with TPU material to provide an upward corrective force, support the arch, and reduce excessive pronation. The second loading region was located on the walls of the heel cup, on the medial and lateral aspects of the heel; in this region, the heel cup and rigid side walls were designed to increase the contact area, thereby stabilizing the calcaneus and maintaining its vertical alignment.

In Parts 2 and 3, the corrected foot postures were collected following the procedure established in Part 1. The FO design was created using Blender (Version 4.3.2, Blender Foundation, Netherlands). The insoles for the running task were fabricated from TPU with a hardness of 60 Shore C; this hardness setting was determined with reference to previous studies[101,102]. Regarding the landing task, the hardness was set to 50 Shore C. This selection was based on a comprehensive assessment of suitable hardness levels identified in previous landing-related research[103,104]. Additional design parameters were implemented as follows: For male subjects, the medial longitudinal arch height was 23.49 ± 2.43 mm, and the heel cup depth was 18.26 ± 2.06 mm. For female subjects, the medial longitudinal arch height was 21.04 ± 2.63 mm, and the heel cup depth was 16.40 ± 2.25 mm. The FO illustrated in **Figure 13B**.

In Part 2, the kinesio tape selected for our study was Kinesio® Tex Gold (Kinesio® Tex Gold™ FP, USA), which had been widely used in previous studies, as illustrated in **Figure 13C**[105]. In the prone position with knees flexed, three strips of kinesio tape were

applied at 100% tension over the middle to improve static stability. An additional strip was applied at 75% tension along the tibialis posterior, from the fifth metatarsal head, around the navicular and medial malleolus, to the mid-tibia. All tapes were rubbed after application to enhance adhesion[19].

In Part 4, designed two different unit cell structures, namely a “#” structure with a 90° angle and an auxetic structure composed of four semi-circular shapes, as illustrated in **Figure 13D**. Based on the plantar pressure distribution of participants, varying material thickness gradients (1/1.5/2/2.5/3mm) were implemented in the high-pressure zones, while the rest of the areas maintained a thickness of 2mm. The unit cell designs and combinations were performed using nTopology 4.1.3 (New York, USA), and smoothed using Geomagic Studio (Geomagic, Inc., Research Triangle Park, NC, United States). The final simulations were conducted using Ansys Workbench 2021 R1 (developed by ANSYS, Inc., located in Canonsburg, Pennsylvania, USA).

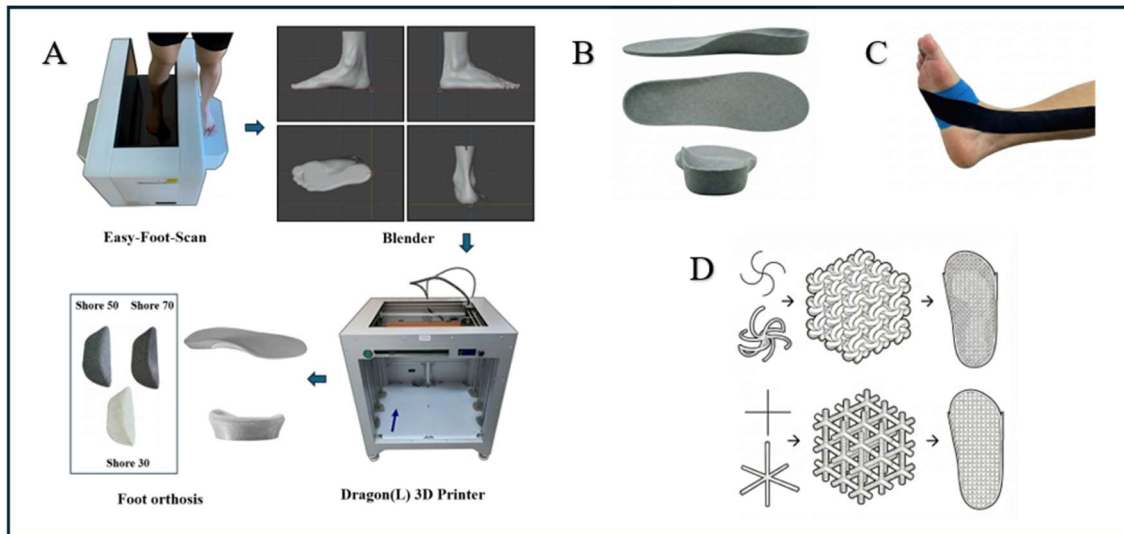


Figure 13 A: FO design and production for walking task; B: FO for running and landing tasks; C: kinesio tape for running task; D: different unit cell structures

2.5 Procedures and data collection

In part 1, based on gait 2392 and Maharaj's foot model[34], 72 reflective markers (16 mm) were placed on bony landmarks by single participant to ensure consistency. Maharaj's foot model had 7 degrees of freedom: ankle (talus–tibia): 3 DoF; subtalar

(calcaneus–talus): 1 DoF; midtarsal (midfoot–calcaneus): 1 DoF; tarsometatarsal (forefoot–midfoot): 1 DoF; and metatarsophalangeal (digits–forefoot): 1 DoF. We started from the generic gait2392 model and supplemented an additional multi-segment foot model, while retaining all original segments and joint definitions of gait2392. Rebecca's research showed that holes under 27 mm don't compromise shoe integrity, and 25 mm apertures don't restrict marker movement[106]. Thus, we made sub-25 mm holes in the shoes to attach markers directly to the feet for accurate motion capture. The experiment was conducted at Ningbo University's Sports Science Laboratory (**Figure 14**).

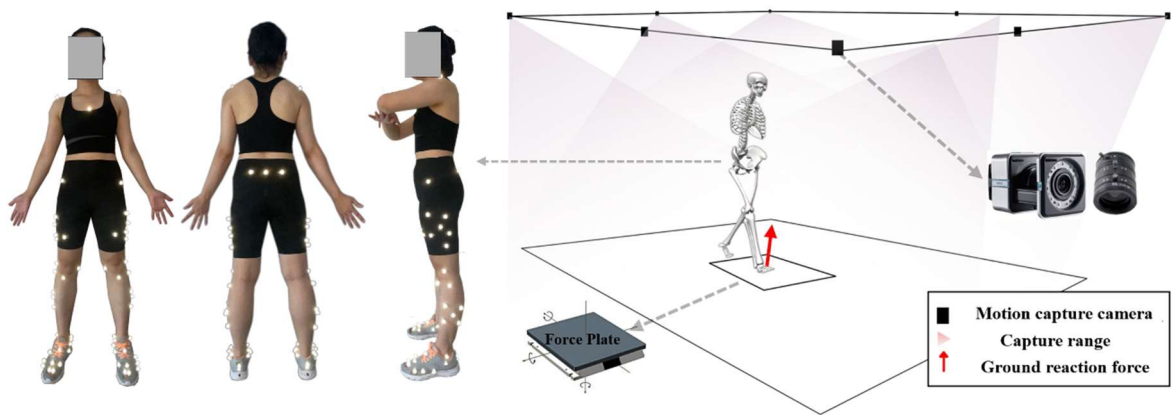


Figure 14 Marker placement and data collection process for walking task

Before data collection, participants determined their most comfortable walking speed and acclimated to it. They wore four FOs: Control Group (CG, no arch support), Hardness A (HA, Shore C 30), Hardness B (HB, Shore C 50), and Hardness C (HC, Shore C 70). Participants walked at their self-selected speed over a force plate (Kistler, Switzerland) embedded in a 40-meter runway, with ground reaction force collected at 2000 Hz. Ten infrared motion capture cameras (Vicon, UK) recorded kinematic data at 200 Hz. The Delsys wireless EMG system synchronously captured signals from the vastus medialis (VM), medial gastrocnemius (MG), tibialis posterior (TP), anterior tibialis (AT), soleus (SOL), and peroneus longus (PL) at 2000 Hz. Each participant completed five successful walking trials per FO type, defined by natural foot placement on the force platform at self-selected speed[64]. Initial contact was when ground reaction force exceeded 10 N[107].

In part 2, reflective marker placement followed the gait 2392 model, and 25 mm holes were made in the shoes to allow direct attachment of markers onto the skin for accurate motion capture[108]. Each participant completed tests under three conditions in a randomized order: the control group (CG) wore standard running shoes without arch support; the foot orthosis group (FOG) wore the same shoes but with foot orthosis replacing the original insole; and the kinesio tape group (KTG) wore standard shoes with the additional application of kinesio tape. Participants acclimated to running at 3.3 m/s ($\pm 5\%$), then ran over a force plate (Kistler, Switzerland) set in a 40-m runway. Ground reaction force were collected at 2000 Hz. Kinematics were recorded using ten infrared cameras (Vicon Motion Systems, Oxford, UK) at 200 Hz. Surface EMG signals were collected from vastus lateralis (VL), lateral/medial gastrocnemius (LG/MG), anterior tibialis (AT), soleus (SOL), and peroneus longus (PL) at 2000 Hz. Five valid trials per condition were collected, requiring natural foot contact on the force plate within the specified speed range (**Figure 15**). Initial contact was defined as when ground reaction force exceeded 10 N[109]. One participant whose knee and ankle joint angles and moments were closest to the mean values was selected. This participant underwent clinical CT scanning of the foot and ankle in the prone position using a Siemens scanner (Munich, Germany), with a slice interval of 0.75 mm



Figure 15 Marker placement and data collection process for running and landing task

In part 3, participants performed single-leg drop-landings from three platform heights (20, 40, and 60 cm) in a randomized order. Reflective marker placement and attachment were identical to Part 2. Participants stood on the platform with hands on their waists to restrict upper-limb contribution, then stepped off (no upward jump) and landed

with the test limb on a force plate (Kistler, Switzerland), maintaining balance without the contralateral foot contacting the plate (**Figure 15**)[110]. Ground reaction forces were collected at 2000 Hz, and kinematics were recorded using ten infrared cameras (Vicon Motion Systems, Oxford, UK) at 200 Hz. Five valid trials per height were collected, requiring natural single-foot contact on the force plate without loss of balance. Initial contact was defined as when vertical ground reaction force exceeded 10 N.

In part 4, participant foot postures in a supine position were collected through clinical CT scans (**Figure 16**). The FootScan pressure plate (size: 2*0.4 m, frequency: 480 Hz, RsScan International, Olen, Belgium) was employed to capture plantar pressure data during barefoot walking conditions at self-selected speed, serving as modeling references.

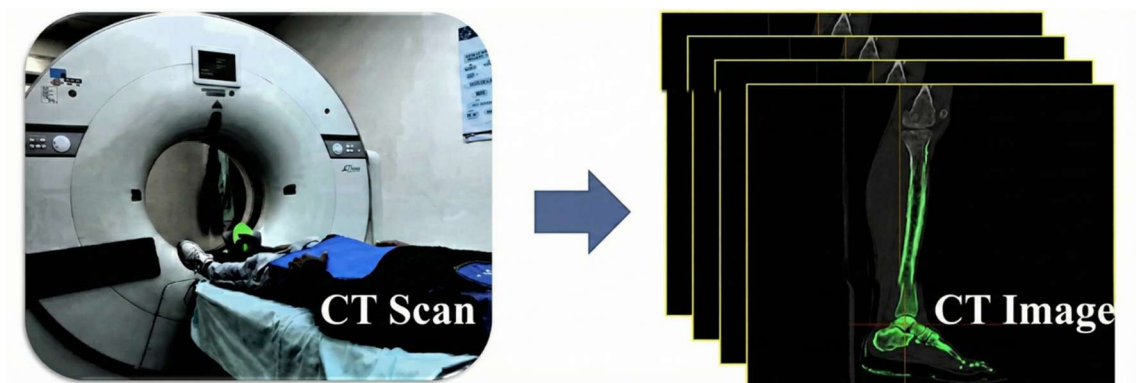


Figure 16 Illustration of CT imaging for the participant

2.6 Data processing

For Parts 1–3, kinematic and GRF were exported from Vicon Nexus (v2.15.0; Vicon Motion Systems, Oxford, UK) as C3D files and processed in MATLAB R2016a (The MathWorks, Natick, MA, USA) for coordinate conversion, filtering, and format conversion. A 4th-order zero-lag Butterworth filter was applied with cut-off frequencies of 6 Hz for marker trajectories and 30 Hz for GRF. Musculoskeletal analyses were performed in OpenSim 4.2 (Stanford University, Stanford, CA, USA) using the gait2392 model for all parts. The Scale tool was used to adjust subject-specific model parameters,

followed by inverse kinematics (IK) and inverse dynamics (ID) to compute joint angles and joint moments.

In Part 1, the gait2392 model was further combined with a multi-segment foot model. After IK and ID, the residual reduction algorithm (RRA) was applied to improve dynamic consistency, computed muscle control (CMC) was used to estimate muscle activations and forces, and joint reaction forces were calculated using the Analysis tool (**Figure 17**). In Part 2, after IK and ID, RRA and CMC were applied to estimate muscle forces; EMG signals were band-pass filtered (10–500 Hz) and then low-pass filtered (10 Hz) for validation[111]. Joint moments, muscle forces, and joint reaction forces (when computed) were normalized to body weight. In Part 3, the OpenSim workflow ended after IK and ID.

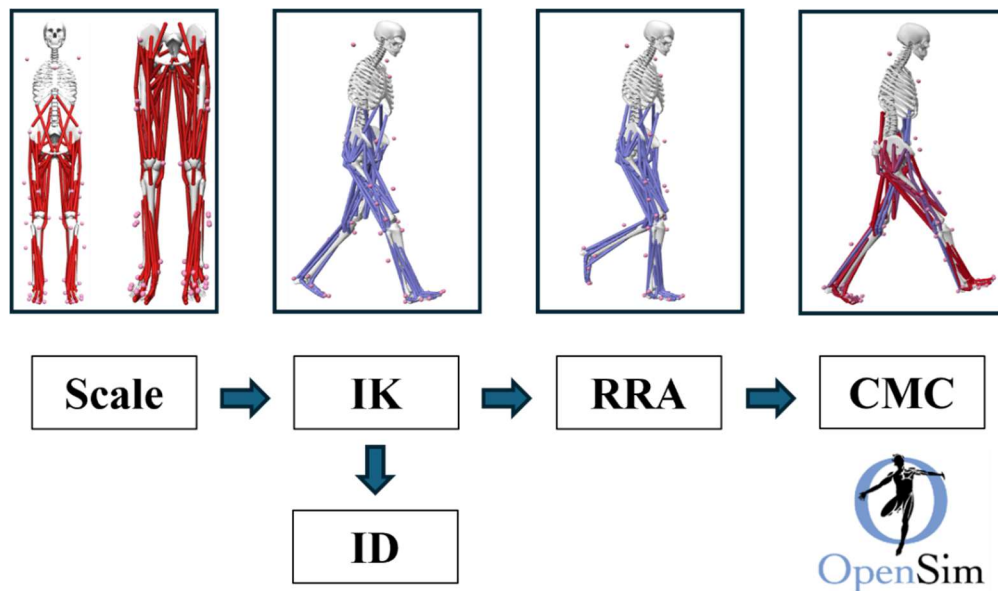


Figure 17 OpenSim computation process

In part 2, we examined peak angles and peak moments in the sagittal and frontal planes of the knee and ankle. Additionally, we assessed joint coordination and its variability in both planes using a modified vector coding technique[112]. For each stance phase (time-normalized to 0–100%), an angle–angle plot was constructed with the proximal joint angle on the x-axis (x) and the distal joint angle on the y-axis (y). The coupling angle γ was defined as the orientation of the vector linking two consecutive samples on this angle–angle plot (i.e., the vector from point i to $i+1$), measured counterclockwise from the positive x-axis. Specifically, for the j th stride at the i th time

point[113], the vector components were $\Delta x = x_{j,i+1} - x_{j,i}$ and $\Delta y = y_{j,i+1} - y_{j,i}$, and the coupling angle was computed as

$$\gamma_{j,i} = \text{atan2}(\Delta y, \Delta x)$$

followed by conversion to degrees and normalization to 0° – 360° (e.g., $\gamma = \gamma \times 180/\pi$; if $\gamma < 0$, then $\gamma = \gamma + 360$). Using atan2 avoids quadrant ambiguity and division-by-zero issues inherent to $\tan^{-1}(\Delta y/\Delta x)$. Here, x and y represent the proximal and distal joint angles, respectively; i denotes the percentage of stance for the j th stride. The coupling angle represents an instantaneous spatial relationship, from which four coordination patterns were identified: (i) in-phase with proximal dominance, (ii) in-phase with distal dominance, (iii) anti-phase with distal dominance, and (iv) anti-phase with proximal dominance (**Figure 18**)[114]. For each participant, coordination variability was calculated as the circular standard deviation of the coupling angle across five strides of data. The coupling angle and variability were calculated using circular statistics[115]. The coordination patterns and variability were categorized according to three stance phases: early (0%-33%), mid (34%- 67%), and late stance (68%-100%)[116].

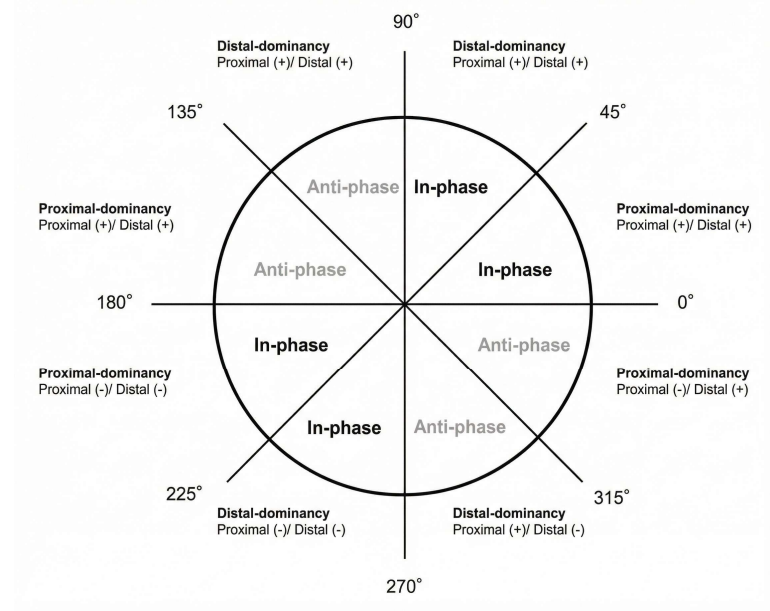


Figure 18 Classification of coordination pattern

We employed CWT to process the ground reaction force (**Figure 19**). According to previous studies[117], the morlet wavelet function was selected to perform in python 3.8

(Python Software Foundation, Wilmington, DE, USA) with a scale of 1–200 (resolution frequency of 0.8125–162.5 Hz) on ground reaction force normalized to vertical (GRFz) initial contact, medial-lateral (GRFx) midstance and anterior-posterior (GRFy) propulsion phase respectively. The indicators of this study included: the mean pseudo-frequency (PFmean), the sum of the signal power (Psum), the maximum power (Pmax), the pseudo-frequency (PFmax), and maximum power percent of stance (%Smax) to reflect the time-frequency domain characteristics[118].

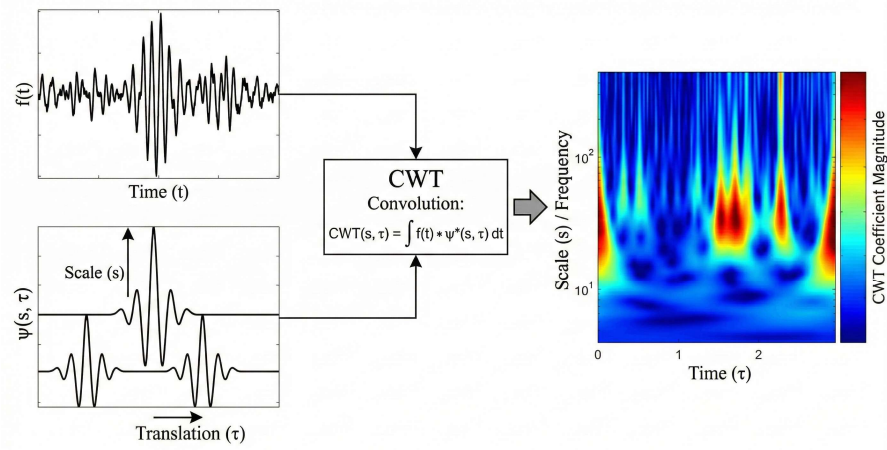


Figure 19 Schematic diagram of CWT

In part 4, Mimics 21.0 software (Materialise, Leuven, Belgium) was utilized to define the boundaries of bones and soft tissues. The STL file of the original foot model was imported into SolidWorks 2021 (SolidWorks Corporation, MA, United States) as a reference to establish the initial insole model. Subsequently, the insole model was partitioned into high-pressure zones based on the plantar pressure data (**Figure 20**).

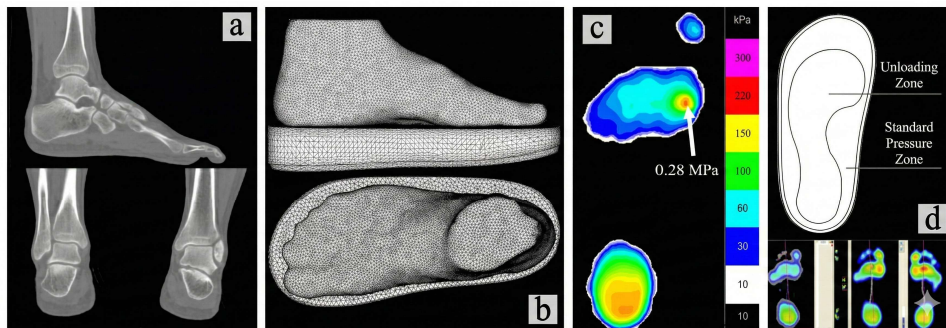


Figure 20 a: Processing CT images with Mimics; b: Establishing the original model based on foot morphology; c: Plantar pressure; d: Insole segmentation

2.7 Statistical analysis

In part 1 and 2, research employed a fully Bayesian statistical framework to derive probabilistic statements (**Figure 21**). Analyses were performed using “noninformative” prior (Cauchy, 0.707)[119]. A one-way repeated measures analysis of variance (ANOVA) was utilized to assess the impact of different interventions. When a meaningful BF_{10} was identified, a Bayesian post-hoc analysis was conducted[120]. Bayesian Pearson correlation analysis was employed to examine the relationships between various biomechanical parameters. The correlation strength ($|r|$) was interpreted using the following: 0-0.3 (weak), 0.3-0.6 (moderate), 0.6-0.9 (strong), and 0.9-1.0 (perfect)[121]. The BF_{10} was interpreted based on the following evidence categories: $< 1/100$ signifies decisive evidence for H_0 ; $1/100$ to $< 1/30$, very strong evidence for H_0 ; $1/30$ to $< 1/10$, strong evidence for H_0 ; $1/10$ to $< 1/3$, substantial evidence for H_0 ; $1/3$ to < 1 , anecdotal evidence for H_0 ; 1 to 3 , anecdotal evidence for H_1 ; >3 to 10 , substantial evidence for H_1 ; >10 to 30 , strong evidence for H_1 ; >30 to 100 , very strong evidence for H_1 ; and >100 , decisive evidence for H_1 . Evidence supporting the alternative hypothesis (H_1) was defined as $BF_{10} > 3$, while evidence for the null hypothesis was defined as $BF_{10} < 1/3$ [122]. All statistical analyses were executed using JASP software (Version 0.9.1, Amsterdam, Netherlands).

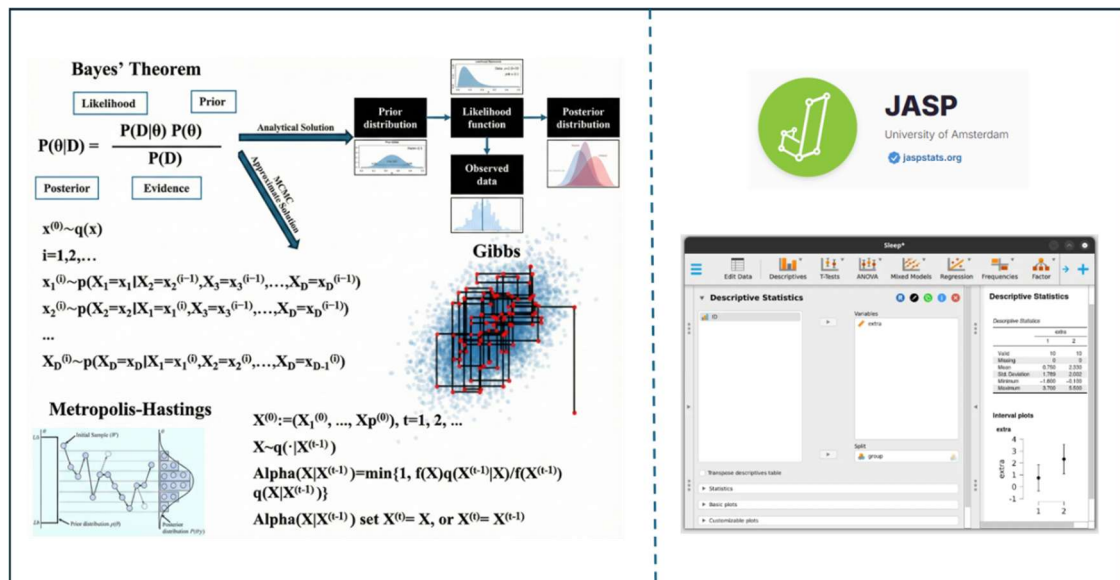


Figure 21 Fundamentals of Bayesian statistics and JASP application

In part 3, statistical calculations were carried out using SPSS version 25.0 software (IBM, Armonk, NY, United States). A one-way repeated-measures ANOVA was used to identify differences in lower-limb kinematic and kinetic parameters during the landing stance phase across three different landing heights. When a significant main effect of landing height was detected, post hoc pairwise comparisons with Holm–Bonferroni correction were performed to account for multiple comparisons. Paired-samples t-tests were used to assess differences between the FO intervention and the non-intervention condition at the same landing height. The level of statistical significance was predefined at $p < 0.05$.

2.8 Foot finite element model and analysis

In part 2, establishment and analysis of the finite element model followed our previous study[123], with a brief description as follows. The grayscale thresholding function in Mimics 21.0 software (Materialise, Leuven, Belgium) was used to define the boundaries of the skin and bone in the foot and ankle, and the foot model was exported as an STL file. The model was then smoothed in Geomagic Studio 2014 (Geomagic, Research Triangle Park, NC, USA). Subsequently, the bones of the foot and ankle were assembled in SolidWorks 2021 (SolidWorks, Waltham, MA, USA) (**Figure 22**). All phalanges were considered as a single rigid body, while cartilage and connective tissues were simulated using 2-mm solid elements. Ligament and fascia attachment sites were determined according to CT images and previous studies. During the simulation process, the mesh convergence error was controlled within 5%. The material parameters were summarized in **Table 1**.

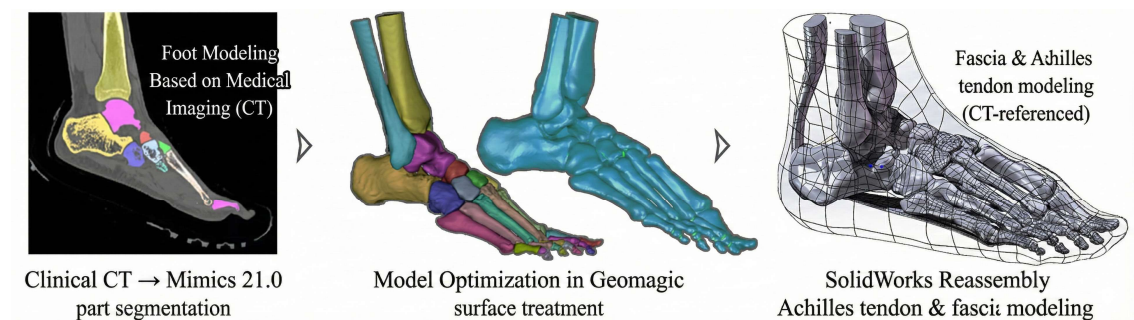


Figure 22 3D Foot Reconstruction and Assembly Pipeline

Table 1 The element types and material properties of the FE model

Component	Type	Property	Density (kg/m ³)	Poisson's ratio ν
Bone	3D-tetrahedra	E: 17000 MPa	1900	0.3
Cartilage		E: 1 MPa	N/A	0.4
Soft Tissue		E: 0.15 MPa	950	0.45
Floor		E: 14200 MPa	1000	0.1
Ligaments	Two-node tension-only truss	Hyperelastic Material Cross-section area: 1.5-5.91 mm ²	1000	N/A
Other Ligaments		Linear Elastic Material E: 264.8 MPa Cross-section area: 1.784 mm ²	1000	0.4
Achilles Tendon	3D-tetrahedra	Hyperelastic Material	1000	N/A
Plantar Fascia			1000	N/A

Note: Ligaments included anterior talofibular, anterior tibiofibular, calcaneofibular, posterior talofibular, tibiocalcaneal, tibiotalar anterior/posterior, spring ligament, and tibionavicular.

Finite element analysis was performed using the transient structural solver in Ansys Workbench (ANSYS, Canonsburg, PA, USA). During the simulation, the upper surfaces of the tibia and fibula were fixed. The degrees of freedom for the primary joints (ankle and metatarsophalangeal joints) were preserved, with contact surfaces set as frictionless to allow relative sliding of the joints. For other joints mediated by cartilage, the contact surfaces between the bone and cartilage were set as bonded contacts to reduce model complexity and computational cost. The coefficient of friction between the skin and the ground was set to 0.532[123]. Ground reaction force obtained from laboratory testing and muscle forces calculated using OpenSim were normalized to a range of 0–100%. To simulate the instant of full-foot contact, five running trials were selected under each of the three conditions where the ankle sagittal plane angle was 0°. The corresponding mean ground reaction force was applied to the ground surface, and the corresponding mean plantarflexor muscle force was applied to the achilles tendon cross-section to simulate muscle contraction (**Figure 23**).

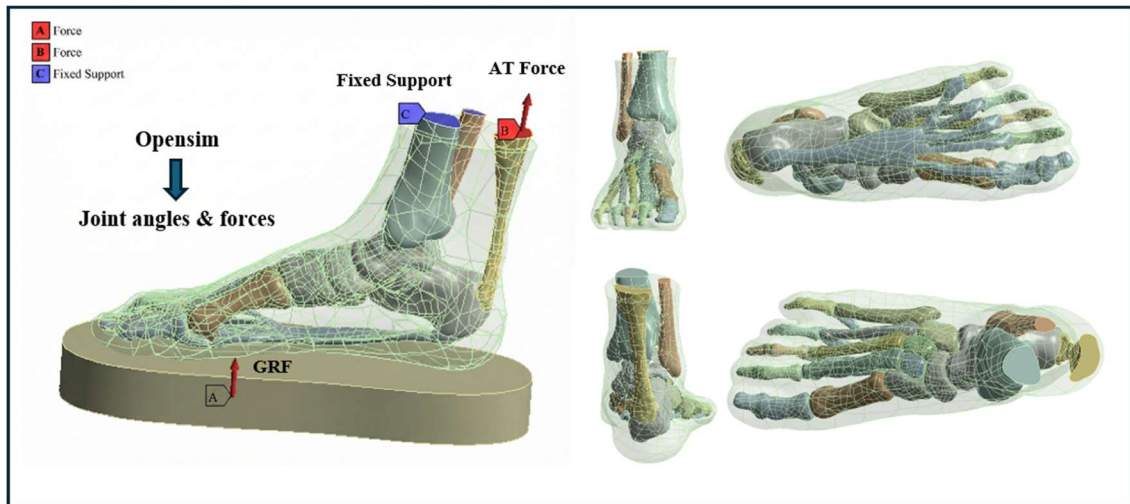


Figure 23 Biomechanical modeling and loading analysis of the human foot

Dynamic biplane X-ray fluoroscopy was recognized as the gold standard for skeletal motion tracking[124]. For model validation, the foot arch compression angle predicted by the FE model was compared with the spatial data obtained from a dual fluoroscopic imaging system (DFIS, Taoying Imaging, Shanghai, China) (**Figure 24**), thereby evaluating the consistency between the simulation results and the experimental data.

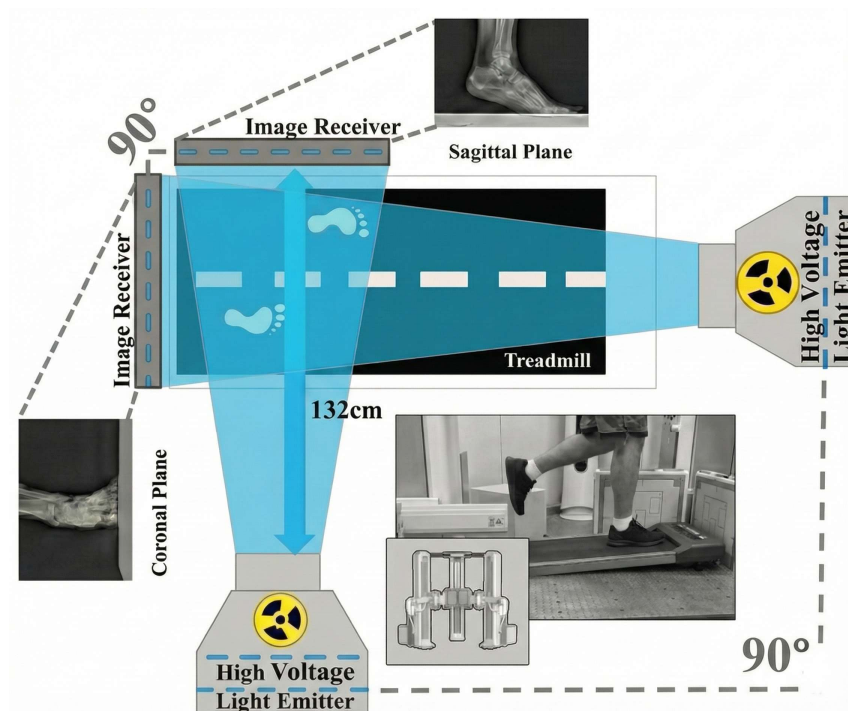


Figure 24 Dual fluoroscopic imaging system

In part 4, the material properties are presented in **Table 2**. Due to the considerably higher stiffness of bones compared to other soft tissues, their influence in this study is minimal, and there is no distinction made between cortical and trabecular bone.

Table 2 Material Properties of Various Components in the Finite Element Model

Material	Young's modulus E (MPa)	Poisson's ratio ν
Bones	14200.00	0.30
Soft tissue	1.15	0.49
Support plate (ground)	17000.00	0.10
Insoles (TPU, Shore A85)	20.00	0.31

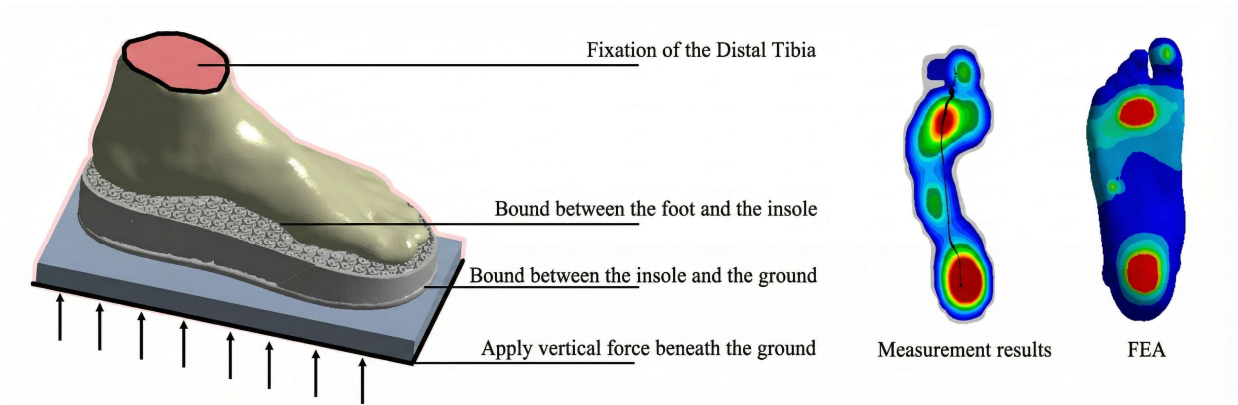


Figure 25 Left, FE boundary conditions: the distal tibia was fixed and a vertical compressive load was applied via the ground plate (deformation solved, not prescribed);
Right, Foot pressure

As depicted in **Figure 25**, the FE model was solved as a quasi-static contact problem. The distal tibia/fibula was fully constrained (all translational and rotational degrees of freedom set to zero) to represent fixation of the shank during stance. A compressive vertical load (750–1350 N, step 150 N) was applied to the plantar-side loading interface via the ground/support plate to emulate different body-weight-related GRF magnitudes. No final displacement or “target position” was prescribed; the resulting deformation and plantar pressure distribution were obtained from the FE solution under the specified constraints and loads. Five simulations were conducted for each of the 2 lattice structures, 5 thickness, and distinct body weight scenarios. The obtained FEA results under barefoot

conditions were compared with experimentally measured plantar pressure data. The similarity in pressure distribution, suggested the effectiveness of the model. Following computation, the average of the maximum peak pressures under each body weight condition was calculated. The insole's capacity to alleviate pressure was assessed as a percentage reduction in pressure (relative to barefoot conditions). Subsequently, scale of the pressure cloud maps was unified, and a total of 250 sets of plantar pressure distribution cloud maps of 2 (structure) * 5 (thickness) * 5 (body weight) * 5 (number of simulations) were exported and cropped to a unified size for subsequent training and validation of algorithm.

2.9 Machine learning algorithm

Machine learning algorithms were used to investigate the relationship between ground reaction forces (hereafter referred to as weight for convenience), insole structure, high-pressure zone thickness and plantar pressure. In this study, we used the plantar pressure obtained from the simulation and the weight applied during the simulation to classify the insole structure and the thickness of the high-pressure zone of the insole, with the aim of recommending the appropriate insole structure and material thickness based on the weight. The data set was divided randomly, with 80% allocated to the training set and 20% to the validation set, ensuring a balanced distribution of samples across all categories. The cloud map of the plantar pressure distribution can be viewed as a 2D image data $P_{a,b}$ with a row * b columns, whose features are extracted using convolutional neural network (CNN), where $a \in [1,2,...,n]$ and $b \in [1,2,...,n]$ represent the rows and columns of the plantar pressure data[125].

The pressure features are extracted using the following equation:

$$f_p = w_p \cdot \text{flatten}(e_{\text{CNN}}(p)) + b_p$$

Where flatten and e are used for neural computation to convert 2D array matrix to 1D array vector and CNN based computation to process the pressure data, w_p and b_p are used to convert the features into weights and biases of desired size. As shown in Figure 4, the model consists of 4 convolutional layers, 4 maximum pooling layers and 2 fully

connected layers. The 4 convolutional layers use different kernel sizes to generate the spatial feature maps of plantar pressure, and the sliding step of the convolutional kernel is 1. The maximum pooling layer reduces the data dimensions to reduce the computational effort, reduce the risk of overfitting and improve the robustness of the model. The Softmax activation function was then used for multi-class classification, the Adam optimisation algorithm was chosen to minimise the loss function, and the validation set was used to assess the accuracy of the model on unseen data (**Figure 26**).

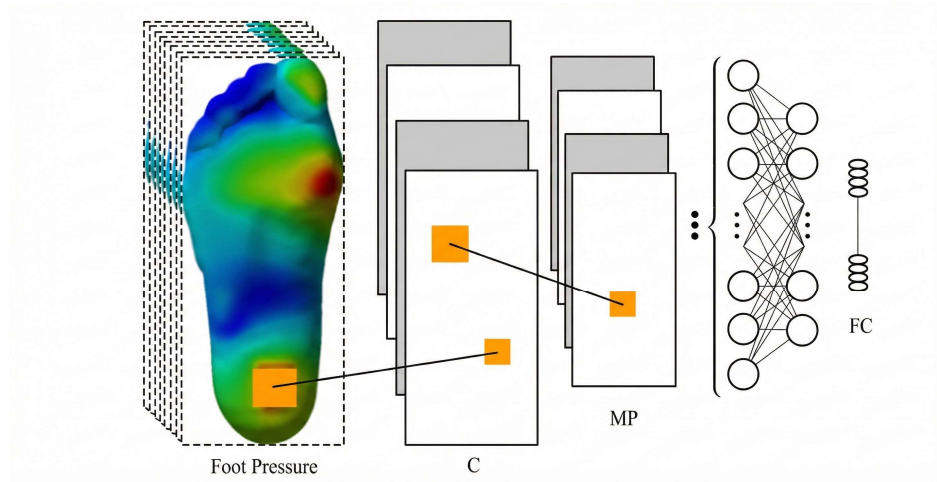


Figure 26 CNN for plantar pressure detection. Where C, MP and FC are the convolutional, maximum pooling and fully connected layers

3 RESULTS

3.1 Results for impact of different arch hardness on walking task

Table 3 and **Table 4** presented ROM, PM, PJRF and PMF (Mean $\pm$ SD) under different conditions. **Figure 27** presented the BF₁₀ values for comparisons across different conditions.

Table 3 ROM, PM and PJRF of lower limb joints under different conditions

Joint Plane	Parameter	CG	HA	HB	HC	BF _{incl}
Knee Sagittal	ROM (°)	49.23$\pm$7.60	55.11$\pm$7.43	52.06 $\pm$ 6.08	49.02$\pm$5.65	31.882
	PM (Nm/kg)	1.26 $\pm$ 0.46	1.35 $\pm$ 0.48	1.36 $\pm$ 0.44	1.36 $\pm$ 0.49	0.607
	PJRF (N/kg)	4.20$\pm$0.82	4.31 $\pm$ 0.86	4.46 $\pm$ 0.98	5.01$\pm$0.92	18.288

Knee Coronal	ROM (°)	4.34±1.23	4.45±1.55	4.84±0.87	5.34±1.20	24.725
	PM (Nm/kg)	0.60±0.12	0.72±0.21	0.69±0.14	0.75±0.17	24.185
	PJRF (N/kg)	2.62±0.51	2.71±0.56	2.64±0.58	2.64±0.61	2.673
Knee Transverse	ROM (°)	17.54±6.63	17.34±4.39	17.26±4.26	17.18±4.89	0.425
	PM (Nm/kg)	-0.14±0.03	-0.12±0.03	-0.13±0.03	-0.14±0.03	2.226
	PJRF (N/kg)	-0.88±0.18	-0.77±0.16	-0.79±0.18	-0.82±0.16	2.609
Ankle Sagittal	ROM (°)	34.35±4.82	35.80±6.29	35.20±4.54	34.59±4.93	0.408
	PM (Nm/kg)	-1.61±0.28	-1.63±0.26	-1.59±0.26	-1.54±0.31	0.491
	PJRF (N/kg)	-4.21±0.79	-4.27±0.86	-4.13±0.75	-4.06±0.77	2.188
Ankle Coronal	ROM (°)	4.90±1.12	3.86±0.96	3.55±0.88	4.11±0.79	30.525
	PM (Nm/kg)	-0.24±0.09	-0.14±0.06	-0.14±0.06	-0.15±0.07	14.853
	PJRF (N/kg)	-1.94±0.40	-1.79±0.50	-1.53±0.46	-1.71±0.52	16.28
Ankle Transverse	ROM (°)	5.83±1.63	6.11±1.89	6.02±1.77	5.98±2.11	1.918
	PM (Nm/kg)	-0.12±0.03	-0.13±0.04	-0.13±0.04	-0.14±0.04	1.291
	PJRF (N/kg)	-0.38±0.11	-0.34±0.09	-0.33±0.08	-0.31±0.09	2.12
Subtalar	ROM (°)	17.96±4.65	14.58±5.97	14.27±5.18	13.96±6.21	5.327
	PM (Nm/kg)	0.19±0.05	0.13±0.05	0.14±0.05	0.14±0.06	23.861
	PJRF (N/kg)	1.26±0.22	1.05±0.24	1.03±0.26	0.96±0.31	11.38
Midtarsal	ROM (°)	12.11±2.62	10.40±2.86	10.66±2.74	10.98±2.45	56.348
	PM (Nm/kg)	-0.31±0.06	-0.33±0.07	-0.33±0.06	-0.34±0.07	0.960
Tarsometatarsal	ROM (°)	7.52±1.67	6.85±1.88	6.85±1.45	6.09±2.11	10.885
	PM (Nm/kg)	-0.27±0.05	-0.29±0.06	-0.28±0.05	-0.28±0.06	2.939
MTP	ROM (°)	18.51±4.89	24.46±4.68	25.70±6.31	26.94±5.86	7.432
	PM (Nm/kg)	0.07±0.01	0.12±0.01	0.11±0.01	0.10±0.01	13.107
	PJRF (N/kg)	2.08±0.41	2.33±0.52	2.38±0.45	2.57±0.44	22.712

Note: Evidence supporting the alternative hypothesis was presented in bold. CG: control group; HA: hardness A; HB: hardness B; HC: hardness C; ROM: range of motion; PM: peak moment; PJRF: peak joint reaction force; BF_{incl}: inclusion bayes factors.

Knee sagittal ROM showed strong evidence, with greater ROM in HA compared to CG and HC (HA–CG: 95% HDI [5.47, 7.25]; HA–HC: 95% HDI [5.08, 7.39]). Knee

coronal ROM showed strong evidence, with greater ROM in HB and HC compared to CG and HC > HB (substantial evidence) (HB–CG: 95% HDI [0.31, 0.88]; HC–CG: 95% HDI [0.82, 1.44] ; HC–HB: 95% HDI [0.39, 0.71]). Ankle coronal ROM showed very strong evidence, with smaller ROM in HA, HB, and HC compared to CG and HB < HC (strong evidence) (HA–CG: 95% HDI [-0.88, -1.27]; HB–CG: 95% HDI [-0.90, -1.69] ; HC–CG: 95% HDI [-0.69, -1.14] ; HC–HB: 95% HDI [-0.32, -0.91]). Subtalar ROM showed substantial evidence, with smaller ROM in HA, HB, and HC compared to CG (HA–CG: 95% HDI [-2.68, -4.22]; HB–CG: 95% HDI [-2.57, -4.69] ; HC–CG: 95% HDI [-3.64, -4.98]). Midtarsal ROM showed very strong evidence, with smaller ROM in HA, HB, and HC compared to CG (HA–CG: 95% HDI [-0.97, -2.22]; HB–CG: 95% HDI [-0.88, -2.03] ; HC–CG: 95% HDI [-0.74, -1.99]). Tarsometatarsal ROM showed strong evidence, with greater ROM in CG compared to HC (CG–HC: 95% HDI [0.85, 1.99]). Metatarsophalangeal (MTP) ROM showed substantial evidence, with greater ROM in HA and HB (strong evidence), and HC compared to CG (HA–CG: 95% HDI [4.54, 7.44]; HB–CG: 95% HDI [4.31, 8.03] ; HC–CG: 95% HDI [4.95, 8.78]).

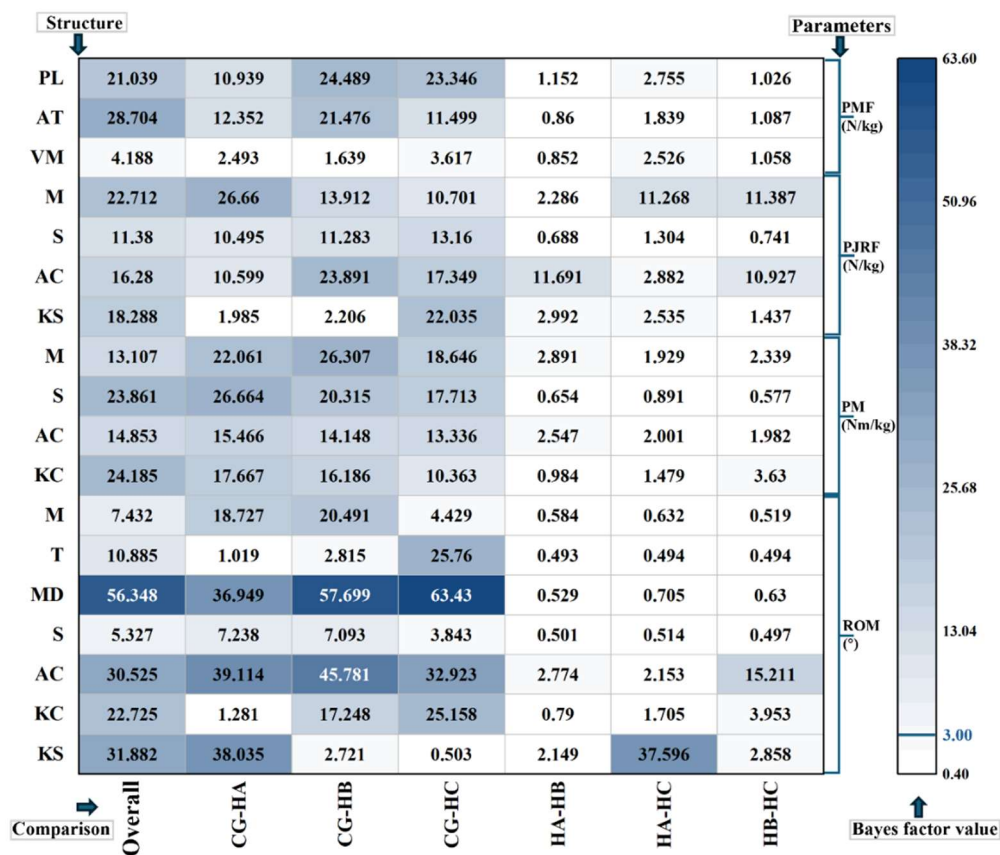


Figure 27 BF_{10} values under different comparative conditions. Darker blue indicated larger Bayes factors, whereas lighter blue indicated smaller values. The x-axis showed comparisons between different conditions, and the y-axis showed the biomechanical parameters included in the comparisons. The color bar on the blue illustrated the mapping between color and Bayes factor, with blue intensity increasing from bottom to top as the Bayes factor value increased; the color corresponding to $BF_{10} = 3$ was marked. KS: knee sagittal; KC: knee coronal; AC: ankle coronal; S: subtalar; MD: midtarsal; T: tarsometatarsal; M: metatarsophalangeal; VM: vastus medialis; AT: anterior tibialis; PL: peroneus longus; ROM: range of motion; PM: peak moment; PJRF: peak joint reaction force; PMF: peak muscle force.

Knee coronal PM showed strong evidence, with greater PM in HA, HB and HC compared to CG and HC > HB (substantial evidence) (HA–CG: 95% HDI [0.03, 0.31]; HB–CG: 95% HDI [0.04, 0.21] ; HC–CG: 95% HDI [0.09, 0.37] ; HC–HB: 95% HDI [0.02, 0.18]). Ankle coronal PM and subtalar PM showed strong evidence, with smaller PM in HA, HB and HC compared to CG (HA–CG: 95% HDI [-0.07, -0.20] [-0.03, -0.10]; HB–CG: 95% HDI [-0.06, -0.21] [-0.04, -0.09]; HC–CG: 95% HDI [-0.06, -0.22] [-0.02, -0.11]). MTP PM showed strong evidence, with greater PM in HA, HB and HC compared to CG (HA–CG: 95% HDI [0.02, 0.10]; HB–CG: 95% HDI [0.01, 0.09] ; HC–CG: 95% HDI [0.01, 0.08]).

Knee sagittal PJRF showed strong evidence, with greater PJRF in HC compared to CG (HC–CG: 95% HDI [0.55, 1.38]). Ankle coronal PJRF showed strong evidence, with smaller PJRF in HA, HB and HC compared to CG, and HB < HA, HC (HA–CG: 95% HDI [-0.08, -0.23]; HB–CG: 95% HDI [-0.21, -0.61] ; HC–CG: 95% HDI [-0.12, -0.48] ; HB–HA: 95% HDI [-0.12, -0.38] ; HB–HC: 95% HDI [-0.15, -0.40]). Subtalar PJRF showed strong evidence, with smaller PJRF in HA, HB and HC compared to CG (HA–CG: 95% HDI [-0.12, -0.38]; HB–CG: 95% HDI [-0.11, -0.49] ; HC–CG: 95% HDI [-0.10, -0.38]). MTP PJRF showed strong evidence, with greater PJRF in HA, HB and HC compared to CG, and HC > HA, HB (HA–CG: 95% HDI [0.18, 0.43]; HB–CG: 95% HDI

[0.17, 0.51] ; HC–CG: 95% HDI [0.31, 0.68] ; HC–HA: 95% HDI [0.12, 0.30] ; HC–HB: 95% HDI [0.11, 0.32]).

Figure 28 presented muscle activations predicted by OpenSim CMC with the experimental EMG trends suggested reasonable qualitative agreement in the timing of the main activation bursts. VM PMF showed substantial evidence, with greater PMF in HC compared to CG (HC–CG: 95% HDI [0.21, 0.48]). AT PMF showed strong evidence, with smaller PMF in HA, HB and HC compared to CG (HA–CG: 95% HDI [-0.08, -0.22]; HB–CG: 95% HDI [-0.07, -0.29] ; HC–CG: 95% HDI [-0.05, -0.28]). PL PMF showed strong evidence, with greater PMF in HA, HB and HC compared to CG (HA–CG: 95% HDI [0.08, 0.33]; HB–CG: 95% HDI [0.17, 0.49]; HC–CG: 95% HDI [0.15, 0.38]). TP PMF showed strong evidence, with smaller PMF in HA, HB and HC compared to CG, and HB > HA (HA–CG: 95% HDI [-0.15, -0.32]; HB–CG: 95% HDI [-0.17, -0.50]; HC–CG: 95% HDI [-0.19, -0.40]; HB–HA: 95% HDI [-0.07, -0.25]).

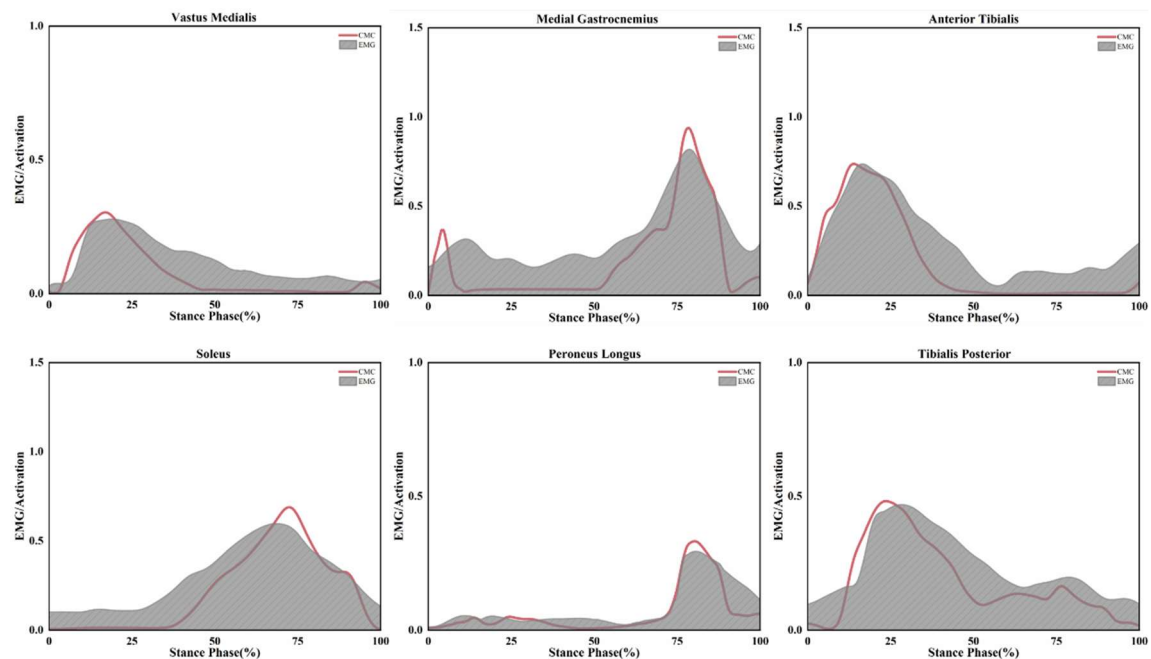


Figure 28 Illustration of the EMG/activation of each muscle. The scale on the left of each image represents that 0 (no activity) to 1 (full activity). The scale on each image below represents a stance phase from 0 to 100. Grey shaded areas represent the results of activation from EMG. Red lines represent the results of activation from CMC.

Table 4 PMF of related lower limb muscles under different conditions

PMF (N/kg)	CG	HA	HB	HC	BF _{incl}
VM	1.85±0.52	1.96±0.57	1.94±0.64	2.21±0.69	4.188
MG	2.11±0.70	2.08±0.67	2.07±0.62	2.06±0.61	0.424
AT	0.55±0.16	0.42±0.13	0.41±0.12	0.40±0.13	28.704
SOL	1.94±0.69	1.92±0.75	1.99±0.80	1.95±0.77	1.463
PL	0.79±0.15	0.97±0.21	1.06±0.23	1.03±0.22	21.039
TP	0.95±0.22	0.73±0.18	0.58±0.20	0.62±0.21	16.255

Note: Evidence supporting the alternative hypothesis was presented in bold. CG: control group; HA: hardness A; HB: hardness B; HC: hardness C; VM: vastus medialis; MG: medial gastrocnemius; AT: anterior tibialis; SOL: soleus; PL: peroneus longus; TP: tibialis posterior; PMF: peak muscle force; BF_{incl}: inclusion bayes factors.

3.5 Correlation analysis

Figure 29 presented the correlation coefficient between biomechanical parameters, revealing several strong associations. Knee coronal ROM and knee coronal PM showed strong positive correlation ($r = 0.68$), indicating strong evidence (BF₁₀: 15.887, 95% HDI [0.61, 0.75]). Ankle coronal ROM and ankle coronal PJRF showed strong positive correlation ($r = 0.66$), indicating strong evidence (BF₁₀: 13.092, 95% HDI [0.60, 0.76]). Ankle coronal ROM and PL PMF showed strong negative correlation ($r = -0.68$), indicating strong evidence (BF₁₀: 17.952, 95% HDI [-0.62, -0.78]). Subtalar ROM and subtalar PJRF showed strong positive correlation ($r = 0.62$), indicating strong evidence (BF₁₀: 18.756, 95% HDI [0.59, 0.79]). Subtalar ROM and PL PMF showed strong negative correlation ($r = -0.69$), indicating strong evidence (BF₁₀: 19.412, 95% HDI [-0.60, -0.81]). MTP ROM and MTP PJRF showed strong positive correlation ($r = 0.69$), indicating strong evidence (BF₁₀: 17.003, 95% HDI [0.58, 0.84]). Knee coronal PM and ankle coronal PM showed strong negative correlation ($r = -0.65$), indicating strong evidence (BF₁₀: 14.339, 95% HDI [-0.57, -0.74]). Knee sagittal PJRF and VM PMF showed strong negative correlation ($r = 0.63$), indicating strong evidence (BF₁₀: 18.874, 95% HDI [0.55, 0.70]).

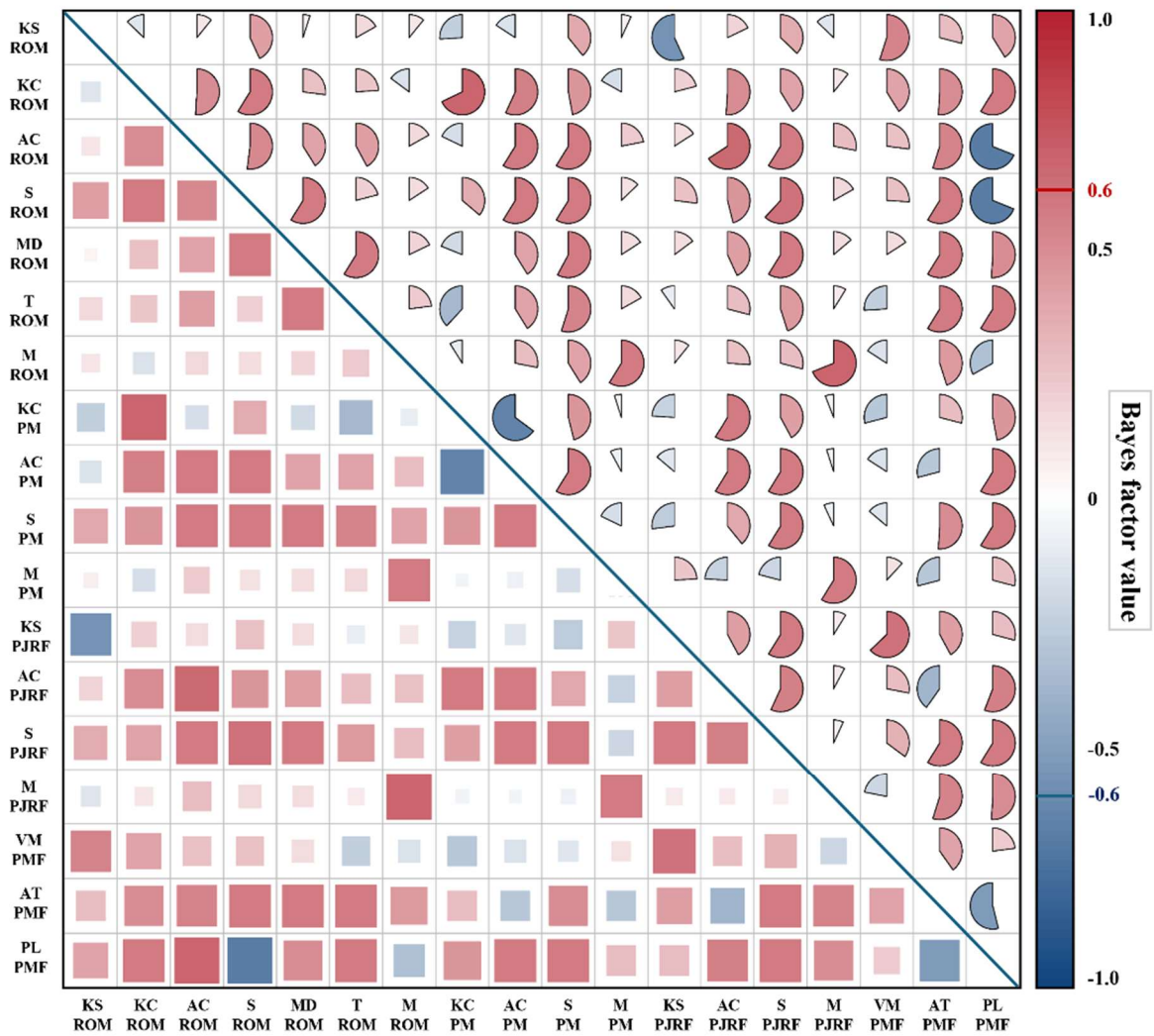


Figure 29 Bayesian correlation between different biomechanical parameters. Blue indicated negative correlations, and red indicated positive correlations. Larger blue squares or circles represented stronger negative correlations, whereas larger red squares or circles represented stronger positive correlations. The x- and y-axes denoted different biomechanical parameters, and, for each x–y parameter pair, the corresponding point represented the magnitude of the correlation between them. The color bar illustrated the relationship between color and the correlation coefficients and marked the color corresponding to $|r|$ values greater than 0.6. KS: knee sagittal; KC: knee coronal; AC: ankle coronal; S: subtalar; MD: midtarsal; T: tarsometatarsal; M: metatarsophalangeal; VM: vastus medialis; AT: anterior tibialis; PL: peroneus longus; ROM: range of motion; PM: peak moment; PJRF: peak joint reaction force; PMF: peak muscle force.

3.2 Results for FO versus kinesio tape on running task

Peak knee adduction angle and moment showed strong evidence, with higher angle and moment in FOG compared to CG and KTG. Peak knee abduction angle showed substantial evidence, with lower angle in FOG compared to CG and KTG. Peak ankle dorsiflexion angle and moment showed substantial evidence, with lower angle and moment in FOG compared to CG. Peak ankle plantarflexion angle showed strong evidence, with lower angle in FOG and KTG compared to CG. Peak ankle eversion angle showed strong evidence, with lower angle in FOG and KTG compared to CG and lower angle in FOG compared to KTG. Peak ankle inversion angle showed substantial evidence, with higher angle in FOG and KTG compared to CG. Peak knee extension moment showed substantial evidence, with higher moment in FOG compared to CG and KTG. Peak knee adduction moment showed substantial evidence, with higher moment in FOG compared to CG and KTG (**Table 5**).

Table 5 Peak angles and moments under different conditions

Parameters		CG	FOG	KTG
Angle(°)	Peak knee flexion	-50.21±1.12	-48.91±1.69	-49.64±1.83
	Peak knee extension	-24.34±1.36	-22.96±1.70	-23.39±0.76
	Peak knee adduction	3.70±0.79#	5.97±0.52#*	4.10±0.14*
	Peak knee abduction	-3.62±0.44#	-2.10±1.09#*	-3.29±0.33*
	Peak ankle Dorsiflexion	28.87±1.43#	24.32±1.08#	26.40±1.27
	Peak ankle Plantarflexion	-30.88±2.72#\$	-24.36±6.69#	-26.43±1.05\$
	Peak ankle Eversion	-9.16±0.57#\$	-1.95±0.77#*	-3.94±0.97\$*
	Peak ankle Inversion	6.06±2.58#\$	9.45±0.92#	9.43±1.52\$
Moment (Nm/kg)	Peak knee flexion	2.99±0.15	3.03±0.17	3.10±0.13
	Peak knee extension	-0.19±0.01#	-0.36±0.05#*	-0.21±0.03*
	Peak knee adduction	-0.72±0.09#	-0.91±0.10#*	-0.78±0.04*
	Peak knee abduction	0.05±0.03	0.04±0.03	0.04±0.03
	Peak ankle Dorsiflexion	0.27±0.03#	0.18±0.02#	0.22±0.04
	Peak ankle Plantarflexion	-2.45±0.14	-2.25±0.09	-2.24±0.11

	Peak ankle Eversion	0.48±0.05	0.43±0.07	0.50±0.03
	Peak ankle Inversion	-0.05±0.01	-0.08±0.01	-0.04±0.01

Note: Evidence supporting the alternative hypothesis was presented: # means CG and FOG; \$ means CG and KTG; *means FOG and KTG.

Knee-ankle sagittal coordination variability showed strong evidence in mid stance, with lower in FOG compared to CG. Knee-ankle sagittal coordination variability showed strong evidence in late stance, with lower in FOG and KTG compared to CG. Knee-ankle frontal coordination variability showed strong evidence in early stance, with lower in FOG compared to CG and KTG. Knee-ankle frontal coordination variability showed strong evidence in mid stance, with lower in FOG and KTG compared to CG, and lower in FOG compared to KTG (**Table 6**).

Table 6 Knee-Ankle coordination variability under different conditions

Parameters	Phase	CG	FOG	KTG
Coordination-sagittal	early stance	8.40±3.58	7.63±3.90	7.25±3.60
	mid stance	10.18±4.87#	6.47±3.70#	8.23±3.76
	late stance	11.18±5.04#\$	5.60±6.01#	7.68±4.56\$
Coordination-frontal	early stance	43.67±36.77#	29.59±19.21#	38.59±26.19
	mid stance	57.33±29.78#\$	19.63±15.86#*	37.58±27.71\$*
	late stance	25.10±12.24	23.21±6.25	21.07±9.82

Note: Evidence supporting the alternative hypothesis was presented: # means CG and FOG; \$ means CG and KTG; *means FOG and KTG.

Knee-ankle sagittal coordination pattern showed substantial evidence in mid and late stance, with lower proportion of in-phase with distal dominant in FOG and KTG compared to CG. Knee-ankle frontal coordination pattern showed strong evidence in early stance, with higher proportion of in-phase with distal dominant and lower proportion of anti-phase with distal dominant in FOG compared to CG and KTG. Knee-ankle frontal coordination pattern showed substantial evidence in mid stance, with lower proportion of

anti-phase with distal dominant in FOG and KTG compared to CG, and lower in FOG compared to KTG (**Table 7**).

Table 7 Knee-Ankle coordination patterns under different conditions

Phase	Patterns (%)	CG	FOG	KTG
Sagittal-early stance	Anti-phase, Proximal-dominant	8.23±3.38	7.65±3.34	7.80±3.05
	In-phase, Proximal-dominant	49.41±8.26	53.53±9.22	51.84±8.72
	Anti-phase, Distal-dominant	20.59±6.08	19.41±6.77	20.26±5.97
	In-phase, Distal-dominant	21.77±5.35	19.41±5.63	20.08±5.05
Sagittal-mid stance	Anti-phase, Proximal-dominant	7.60±3.83	6.06±3.06	6.78±2.64
	In-phase, Proximal-dominant	31.51±7.66	37.78±7.42	36.57±7.42
	Anti-phase, Distal-dominant	2.10±1.89	2.42±1.42	2.28±1.51
	In-phase, Distal-dominant	60.00±9.44#\$	52.52±8.84#	54.37±7.16\$
Sagittal-late stance	Anti-phase, Proximal-dominant	22.57±5.42	25.24±4.79	24.32±4.64
	In-phase, Proximal-dominant	3.03±1.78	4.24±1.97	3.78±2.01
	Anti-phase, Distal-dominant	20.08±4.07	23.96±4.98	23.03±5.11
	In-phase, Distal-dominant	54.40±9.45#\$	46.55±9.31#	48.87±8.77\$
Frontal-early stance	In-phase, Proximal-dominant	11.17±3.16	9.47±3.83	10.82±4.02
	Anti-phase, Proximal-dominant	25.88±6.84	28.76±7.29	27.12±6.51
	In-phase, Distal-dominant	3.53±1.46#	17.65±6.07#*	6.99±2.93*
	Anti-phase, Distal-dominant	59.41±8.42#	44.12±9.06#*	55.07±9.25*
Frontal-mid stance	In-phase, Proximal-dominant	3.64±2.42	5.85±2.71	5.15±2.71
	Anti-phase, Proximal-dominant	18.18±6.68	22.88±5.49	20.31±4.49
	In-phase, Distal-dominant	22.42±7.07	27.51±7.44	24.71±8.44
	Anti-phase, Distal-dominant	55.76±9.19#\$	43.76±9.92#*	49.83±10.02\$*
Frontal-late stance	In-phase, Proximal-dominant	20.00±6.98	19.33±5.71	17.99±6.67
	Anti-phase, Proximal-dominant	21.82±5.95	22.48±5.41	23.92±5.90
	In-phase, Distal-dominant	26.06±7.30	23.63±6.45	23.92±6.77
	Anti-phase, Distal-dominant	32.12±7.07	34.54±7.05	34.15±7.57

Note: Evidence supporting the alternative hypothesis was presented: # means CG and FOG; \$ means CG and KTG; *means FOG and KTG.

GRFx midstance %Smax showed substantial evidence, with earlier percent in FOG compared to CG. GRFy propulsion phase Psum showed strong evidence, with lower sum power in KTG compared to CG. GRFy propulsion phase Pmax and %Smax showed substantial evidence, with lower maximum power and later percent in KTG compared to CG (**Table 8**). Time-frequency domain analysis cloud of the ground reaction force was shown as **Figure 30**.

Table 8 Time-frequency analysis results of GRF under different conditions

Phase	Parameters	CG	FOG	KTG
GRFz Initial Contact	PFmean	18.36±1.48	19.22±2.16	17.90±1.10
	Psum	313.81±95.19	409.11±83.16	366.59±71.74
	Pmax	2.32±0.81	2.71±0.61	2.66±0.57
	PFmax	9.43±1.78	11.38±3.40	9.91±1.67
	%Smax	70.14±4.43	73.10±7.77	71.09±1.49
GRFx Midstance	PFmean	7.01±0.97	7.88±2.35	6.47±0.54
	Psum	9.49±4.37	10.08±6.37	6.41±2.72
	Pmax	0.09±0.04	0.10±0.07	0.06±0.03
	PFmax	3.58±0.45	3.41±0.36	3.41±0.36
	%Smax	58.67±1.52#	53.61±2.51#	55.00±3.61
GRFy Propulsion Phase	PFmean	3.38±0.97	3.36±0.59	3.91±0.46
	Psum	56.14±8.25\$	51.34±10.41	40.96±4.49\$
	Pmax	0.46±0.10\$	0.44±0.11	0.32±0.05\$
	PFmax	2.76±0.45	2.76±0.45	2.93±0.45
	%Smax	45.92±3.67\$	51.64±4.44	51.77±1.72\$

Note: Evidence supporting the alternative hypothesis was presented: # means CG and FOG; \$ means CG and KTG; *means FOG and KTG.

Regarding the FEA, the peak stress in the plantar fascia was measured at 7.85 MPa in the CG, decreasing to 6.93 MPa in the KTG, and further reducing to 6.45 MPa in the FOG. In the anterior longitudinal arch (metatarsals), the peak stress measured 6.95 MPa in the CG, decreased to 5.28 MPa in the KTG, and further declined to 4.80 MPa in the

FOG. In the mid-portion of the longitudinal arch (first to third cuneiforms, navicular, and cuboid bones), the stress was 0.17 MPa in the CG, increased to 0.26 MPa in the KTG, and further rose to 0.34 MPa in the FOG. In the posterior longitudinal arch (talus and calcaneus), the stress values were 3.22 MPa in the CG, reduced to 2.99 MPa in the KTG, and further decreased to 2.39 MPa in the FOG (**Figure 30**).

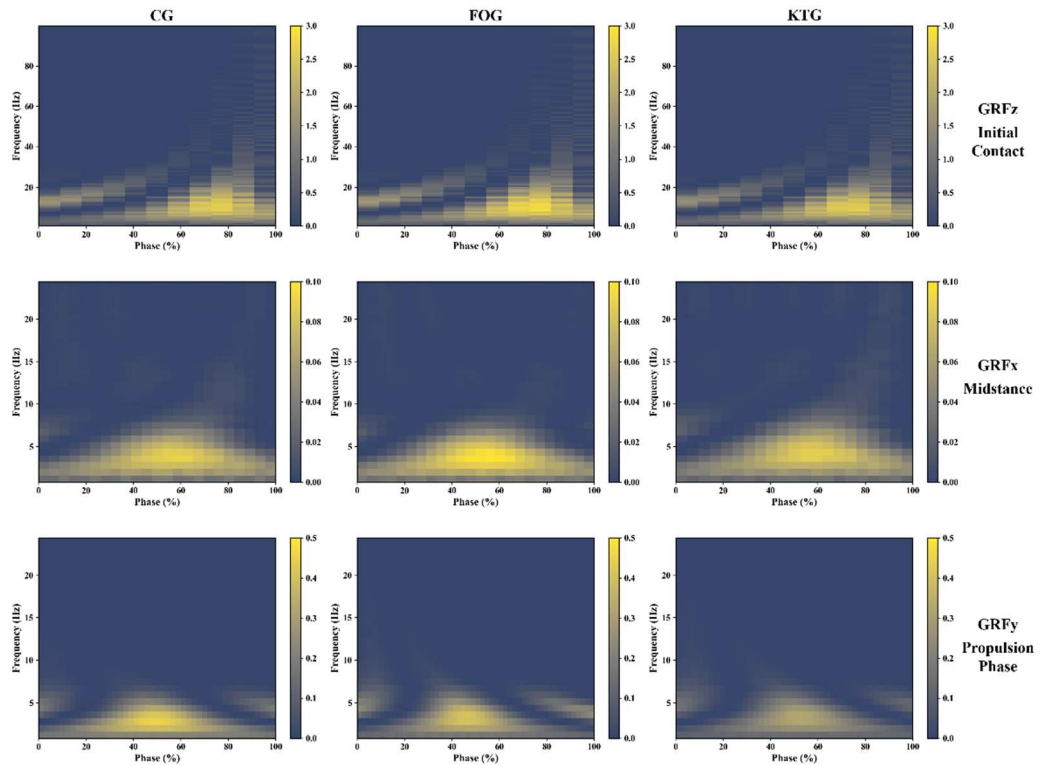


Figure 30 Time-frequency domain analysis cloud of the ground reaction force

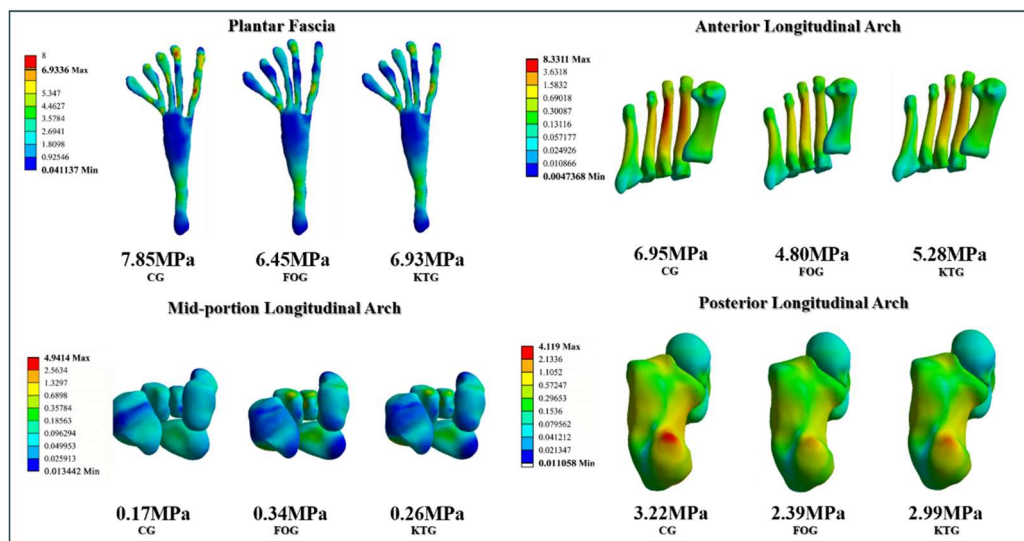


Figure 31 Stress distribution maps of the plantar fascia and longitudinal arch segments

3.3 Results for FO impact on landing task under different heights

Peak angles and moments (hip flexion, knee flexion, ankle dorsiflexion), peak vGRF (forefoot, rearfoot) and maximum LR (forefoot, rearfoot) exhibited significantly greater with landing height in both conditions. Peak angles (hip adduction, hip internal rotation, knee abduction, knee internal rotation, ankle eversion) exhibited significantly smaller under Fo condition in all landing heights. Peak moments (knee abduction, ankle eversion) exhibited significantly smaller under Fo condition in all landing heights. Maximum LR (forefoot, rearfoot) exhibited significantly smaller under Fo condition in all landing heights. Angles and moments change % (knee abduction, ankle eversion) exhibited significantly greater at 60 cm than at 20 cm (**Table 9**).

Table 9 Kinematics and kinetics during landing tasks under different conditions

Variable	Height (cm)	NFO	FO	Change %
Peak angle (°)				
Hip flexion	20	29.33±3.21 ^{ab}	28.67±4.03 ^{AB}	2.25±0.31
	40	39.71±3.62 ^{ac}	38.14±4.68 ^{AC}	3.95±0.41
	60	61.14±2.99 ^{bc}	59.78±3.31 ^{BC}	2.22±0.19
Hip adduction	20	11.56±3.74 [#]	7.71±1.93 [#]	33.3±5.64
	40	12.65±3.00 [#]	8.42±2.25 [#]	33.44±8.16
	60	12.66±3.17 [#]	9.02±2.87 [#]	28.75±8.66
Hip internal rotation	20	17.08±2.99 [#]	13.78±2.66 [#]	19.32±3.98
	40	18.64±2.86 [#]	14.55±2.34 [#]	21.94±3.05
	60	19.44±2.98 [#]	15.06±2.81 [#]	22.53±3.17
Knee flexion	20	50.14±4.78 ^{ab}	49.16±5.47 ^{AB}	1.95±0.84
	40	66.66±5.40 ^{ac}	66.10±4.58 ^{AC}	0.84±0.08
	60	83.11±3.56 ^{bc}	82.59±4.38 ^{BC}	0.63±0.07
Knee abduction	20	4.83±0.87 [#]	3.23±1.06 [#]	33.13±8.13*
	40	5.51±0.91 [#]	3.43±1.30 [#]	37.75±3.77
	60	6.29±1.21 [#]	3.51±1.24 [#]	44.21±11.54*

Knee internal rotation	20	9.93±1.66 [#]	6.74±1.23 [#]	32.12±5.92
	40	10.86±1.52 [#]	7.79±1.01 [#]	28.27±4.30
	60	13.04±1.69 [#]	9.08±1.33 [#]	30.37±3.37
Ankle dorsiflexion	20	18.16±6.08 ^{ab}	16.69±6.16 ^{AB}	8.09±2.58
	40	26.41±6.15 ^{ac}	23.92±8.05 ^{AC}	9.43±3.01
	60	33.33±5.37 ^{bc}	30.95±7.14 ^{BC}	7.14±1.83
Ankle eversion	20	6.04±0.66 [#]	3.99±0.65 [#]	33.94±3.11 [*]
	40	6.95±0.52 [#]	4.12±0.65 [#]	40.72±5.36
	60	7.87±0.81 [#]	4.33±0.64 [#]	44.98±6.27 [*]
Peak moments (Nm/kg)				
Hip flexion	20	2.30±0.81 ^{ab}	1.99±0.65 ^{AB}	13.48±3.18
	40	3.42±0.79 ^{ac}	3.00±0.54 ^{AC}	12.28±2.22
	60	4.56±0.84 ^{bc}	4.09±0.77 ^{BC}	10.31±2.05
Knee flexion	20	1.90±0.39 ^{ab}	1.88±0.54 ^{AB}	1.05±0.24
	40	3.18±0.35 ^{ac}	3.15±0.23 ^{AC}	0.95±0.09
	60	4.33±0.51 ^{bc}	4.22±0.59 ^{BC}	2.54±0.47
Knee abduction	20	0.74±0.26 [#]	0.54±0.21 [#]	27.03±11.50 [*]
	40	0.85±0.19 [#]	0.59±0.27 [#]	30.59±7.36
	60	0.98±0.23 [#]	0.61±0.31 [#]	37.76±11.22 [*]
Ankle dorsiflexion	20	1.24±0.31 ^{ab}	1.19±0.28 ^{AB}	4.03±1.12
	40	2.31±0.25 ^{ac}	2.16±0.19 ^{AC}	6.49±0.74
	60	3.50±0.36 ^{bc}	3.27±0.39 ^{BC}	6.57±0.81
Ankle eversion	20	0.50±0.13 [#]	0.33±0.19 [#]	34.02±10.29 [*]
	40	0.58±0.15 [#]	0.35±0.21 [#]	39.66±12.30
	60	0.68±0.15 [#]	0.37±0.22 [#]	45.59±10.86 [*]
GRF (BW) & LR (BW/s)				
Forefoot peak vGRF	20	2.26±0.24 ^{ab}	2.19±0.33 ^{AB}	3.10±0.47
	40	3.34±0.30 ^{ac}	3.26±0.25 ^{AC}	2.40±0.29

	60	4.51±0.47 ^{bc}	4.43±0.38 ^{BC}	1.77±0.22
Rearfoot peak vGRF	20	0.92±0.08 ^{ab}	0.87±0.11 ^{AB}	5.43±0.67
	40	1.95±0.12 ^{ac}	1.82±0.09 ^{AC}	6.67±0.51
	60	2.74±0.11 ^{bc}	2.65±0.09 ^{BC}	3.28±0.19
Forefoot max LR	20	80.27±22.87 ^{ab#}	63.49±19.79 ^{AB#}	20.90±5.41
	40	127.50±30.08 ^{ac#}	102.24±28.56 ^{AC#}	19.81±4.99
	60	198.73±33.28 ^{bc#}	160.57±27.42 ^{BC#}	19.20±3.62
Rearfoot max LR	20	6.38±3.64 [#]	4.76±2.55 [#]	25.39±12.08
	40	10.96±4.62 [#]	7.46±3.86 [#]	31.93±13.14
	60	14.28±5.07 [#]	10.08±6.35 [#]	29.41±8.63

Notes: ^{abc,ABC} Within each condition, different letters indicated significant differences among landing heights ($p<0.05$). [#]FO significantly differed between FO and NFO at the same landing height ($p<0.05$). *Change % at 60 cm significantly greater than at 20 cm ($p<0.05$).

3.4 Results for customized 3D-printed insoles: FEA and CNN

As depicted in **Figure 32**, when simulating the most suitable insole under corresponding body weight conditions, the average reduction in peak plantar pressure was 39.2% (relative to barefoot conditions). Furthermore, in comparison to the “#” structure, the auxetic structure insoles appear to exhibit superior pressure alleviation performance, i.e., a greater reduction in pressure across different thickness scenarios. It is noteworthy that the reduction in pressure is not necessarily more effective with lower material thickness (i.e., softer). When higher plantar pressure is applied (simulating larger body weight conditions) and the material thickness in the high-pressure region of the insole is greater (2.5mm and 3mm), there seems to be an enhanced pressure reduction effect. Unfortunately, the CNN achieved a classification accuracy of only 60.75% for the insole structure and thickness, indicating a relatively low precision in model classification.

Structure	Thickness	700N	900N	1050N	1200N	1350N	AVE
	1 mm	22.8	28.4	12.6	13.0	11.8	17.7
	1.5 mm	28.6	47.5	15.5	12.7	33.6	27.6
	2 mm	13.2	35.3	29.4	23.6	39.4	28.2
	2.5 mm	11.4	12.2	13.2	48.7	40.5	25.2
	3 mm	15.0	3.2	8.9	36.1	24.1	17.5
	1 mm	28.3	14.6	12.3	16.0	6.9	15.6
	1.5 mm	21.7	22.0	30.9	27.5	22.9	25.0
	2 mm	22.0	38.6	26.1	32.3	20.2	27.8
	2.5 mm	8.6	26.9	6.8	34.7	31.9	21.8
	3 mm	5.9	-4.5	18.7	28.5	38.2	17.4

Figure 32 Pressure reduction effects of auxetic structure insoles and “#” shaped structure insoles of varying thicknesses under different body weight simulations

4 DISCUSSION

4.1 Impact of different arch hardness on walking task

The hypothesis that, as hardness increased, peak tibialis posterior (TP) muscle force would be further reduced was partially confirmed. Under the FO intervention, the muscular demand of TP during the stance phase decreased, and its peak muscle force (PMF) was reduced. However, as hardness further increased, changes in TP PMF tended to plateau, and the unloading effect on TP was no longer obvious. This plateau phenomenon may have reflected two factors: first, the passive arch support that the FO could provide might have approached its upper limit; second, to maintain foot stability and control subtalar joint motion, TP still needed to sustain a certain level of activation[126], which limited the scope for further reductions in PMF.

The hypothesis that knee joint load-related parameters would increase with increasing hardness was confirmed. Under the FO intervention, knee coronal range of motion (ROM) and peak moment (PM) increased, which was consistent with previous research[127,128]. Ankle coronal PM and knee coronal ROM were highly correlated with the increase in knee coronal PM, suggesting that changes in coronal-plane motion and loading at the ankle and knee were the main drivers of the elevated knee coronal PM[88]. Under the relatively high hardness condition (HC condition), knee sagittal peak joint reaction force (PJRF) increased, likely because the harder arch support reduced shock absorption at the foot, transmitting more impact force to the knee. The PMF of the vastus

medialis also increased, likely as an adaptation to the increase in PJRF. These two parameters were highly correlated, and the increased PJRF subjected the knee to greater compressive loading[129].

The hypothesis that ankle and foot biomechanical measures would show further improvement as hardness increased was partially confirmed. Under the FO intervention, ankle coronal ROM, subtalar ROM, and midtarsal ROM all decreased, findings consistent with prior research and likely attributable to reduced eversion and pronation[130,131]. With FO, tibialis anterior (TA) PMF decreased, probably because less overactivity was required to maintain the medial arch and control eversion[132]. By contrast, improved subtalar alignment and reduced pronation optimized the length–tension relationship of the peroneus longus (PL), enhancing its arch-stabilizing and eversion-control capacity and potentially increasing PL PMF[49]. FO reduced ankle coronal and subtalar PM, as well as PJRF, findings consistent with prior research and were probably explained by its ability to limit excessive ROM and joint loading during walking[133-136]. However, as hardness increased, all of these parameters, except ankle coronal ROM, no longer changed. Under the relatively high hardness condition (HC condition), ankle coronal ROM did not further decrease and was instead slightly larger than in the HB condition. This finding indicated that, once hardness reached a certain level, the corrective effects on rearfoot alignment and arch support tended to plateau, and similar nonlinear responses to FO had also been reported in previous research[137,138].

Patients with flatfoot exhibited increased dorsiflexion at the first tarsometatarsal joint[139], and FO may have helped improve this condition[5]. While HA and HB showed limited effects, HC effectively reduced compensatory tarsometatarsal motion, enhanced metatarsal stability, and thereby decreased ROM. FO increased the metatarsal declination angle, placing the first metatarsal in relative plantarflexion and favoring the propulsive phase of gait[140], which may have explained the greater metatarsophalangeal (MTP) ROM. Arch elevation also redistributed plantar pressure toward the forefoot[141], concentrating propulsion at the MTP joint and increasing PM and PJRF. PJRF increased as hardness increased, which was consistent with the notion that enhanced midfoot

support reduced midfoot load while shifting pressure distally to the MTP joint[76]. MTP ROM was strongly and positively correlated with MTP PJRF, indicating that, at the MTP joint, increasing hardness improved its biomechanical performance but at the same time increased joint loading[142].

4.2 FO versus kinesio tape on running task

Based on Bayesian statistical analysis, the present study evaluated the effects of two interventions on knee and ankle biomechanics during running, including changes in joint angles and moments, inter-joint coordination patterns and their variability, time–frequency characteristics of vertical ground reaction forces, and stress distribution in foot bones and fascia. The comprehensive analysis demonstrated that both FOs and kinesio tape produced beneficial biomechanical effects, although biomechanical differences between the two interventions were also observed.

Compared with CG and KTG, FOG showed an increase in peak knee extension moment. Compared with kinesio tape, FO rendered the foot relatively more rigid, reducing its ability to absorb impact. Consequently, the knee joint needed to take on more of the shock absorption to control trunk descent and maintain stability. This indicated that during ground contact, the knee extensors contracted more actively to resist the larger external flexion moment, leading to an increased peak extension moment[18]. Compared with CG and KTG, FOG also demonstrated increased peak knee adduction angle and moment, while peak knee abduction angle decreased. By providing stronger structural support, FO shifted the knee joint into a slightly more adduction position, thereby reducing the peak knee abduction angle but increasing the peak knee adduction angle and moment[143,144]. However, the increase in peak adduction moment might have been a risk factor for knee osteoarthritis[145]. In contrast, kinesio tape, due to its elastic nature and relatively low structural support, had no effect on knee joint angles and moments in the sagittal and frontal planes.

Compared with CG, FOG demonstrated decreased peak ankle dorsiflexion angle and moment. Previous studies showed that runners with flatfoot exhibited arch collapse

during running, resulting in excessive ankle dorsiflexion and eversion, talotarsal dislocation, and pronation[68,146]. FO alleviated these symptoms by reducing ankle dorsiflexion, thereby decreasing the corresponding peak moment. Compared with CG, both FOG and KTG showed reduced peak ankle plantarflexion angle. With medial arch support provided by FO and kinesio tape, the foot was aligned closer to a neutral position, reducing the need for excessive plantarflexion during stance. This led to a more stable foot strike and diminished the demand for forefoot plantar pressure during propulsion[8,147]. Compared with CG, both FOG and KTG reduced the peak ankle eversion angle, with FOG demonstrating a greater reduction; conversely, peak ankle inversion angle increased. Excessive eversion was the most prominent functional abnormality in flatfoot. FO was shown to effectively correct excessive eversion during walking[60], and our results confirmed similar effects during running. To maintain frontal plane mobility, inversion increased slightly as a compensatory response. Kinesio tape also provided some corrective benefit, but due to its lower structural stiffness compared to FO, its effects on frontal and sagittal ankle kinematics and kinetics were less pronounced, with smaller reductions in peak ankle eversion angle and plantarflexion moment.

In the sagittal plane, variability decreased in the mid and late stance phases with FOG, while KTG also demonstrated decreased variability during late stance. Reduced variability might have stemmed from the limited ankle range of motion under FO and kinesio tape, resulting in more consistent coordination[148]. In the frontal plane, variability decreased during early stance due to reduced excessive eversion and pronation following arch support; during mid stance, when pronation peaked in flatfoot, FO limited pronation, leading to more stable coordination. Overall, FO reduced frontal plane coordination variability by restricting ankle motion. Kinesio tape produced similar but less pronounced effects[149]. In sagittal mid stance, the proportion of “In-phase, Distal-dominant” coordination patterns decreased significantly. Specifically, mid-stance coupling typically involved knee flexion and ankle dorsiflexion; by reducing excessive dorsiflexion, arch support decreased reliance on ankle-dominant coupling. Improved knee–ankle coordination suggested a positive adaptation for stabilizing transition during

mid-stance[150]. During sagittal late stance (propulsion phase), the proportion of “In-phase, Distal-dominant” coordination decreased significantly, indicating more coordinated knee–ankle interaction during push-off. This might have been attributed to reduced plantarflexion demand from FO and kinesio tape, resulting in more efficient movement and a stable propulsion strategy [151]. In the frontal plane, FO increased the proportion of “In-phase, Distal-dominant” coordination while reducing “Anti-phase, Distal-dominant” patterns during early stance; both FO and kinesio tape reduced anti-phase patterns during mid-stance. Overall, interventions increased in-phase patterns and reduced anti-phase patterns. Previous studies suggested FO decreased ankle eversion and tibial internal rotation, consistent with our kinematic findings[152]. These changes naturally reduced the occurrence of anti-phase, ankle-dominant coupling. Although kinesio tape did not increase in-phase proportions, it did reduce anti-phase proportions. An increased proportion of in-phase coordination indicated more synchronized frontal plane knee–ankle motion, which might have enhanced lateral stability during early and mid-stance. In-phase motion reduced relative rotation between knee and ankle, improving efficiency of force transfer. Conversely, reduced anti-phase patterns might have lowered the risk of running-related injuries[113].

Compared with CG, the timing of %Smax occurrence in GRFx during midstance occurred earlier in FOG, whereas the Pmax values showed no significant difference. GRFx was related to the maintenance of balance and stability during straight running[153]. This indicated that during the midstance phase, the center of mass was adjusted laterally at an earlier stage, adopting a more efficient postural control strategy. Such earlier but smooth and controllable adjustment might have been beneficial for improving running efficiency. Compared with CG, KTG exhibited reduced Psum and Pmax of GRFy during the propulsion phase, while the timing of %Smax occurrence was delayed. Although the running speed was controlled to be consistent across groups, KTG achieved the same speed with lower Psum and Pmax. Such reductions generally suggested lower plantar impact and potentially greater running efficiency. The delayed %Smax also reflected an adjustment in propulsion strategy: runners might have slowed down the rate of force

application, thereby prolonging the propulsion phase and flattening the originally sharp force impulse. The elastic energy return mechanism of the tendon could have contributed by allowing muscle fibers to contract nearly isometrically, releasing force over a longer period and at a lower velocity, thus reducing muscular energy expenditure and enhancing running efficiency[154]. The relatively better effect of kinesio tape might have been associated with its role in muscle activation and the improvement of the transverse arch. While supporting the longitudinal arch, it also indirectly reduced the width of the transverse arch; the inward force caused the metatarsal heads to converge toward the center, thereby indirectly elevating and supporting the transverse arch. In addition to providing mechanical support, kinesio tape also enhanced sensory input. This increased proprioceptive feedback might have facilitated more effective activation of related foot muscles such as the interossei, lumbricals, and tibialis posterior.

Finite element analysis demonstrated that once the arch was supported, the peak stress on the plantar fascia decreased significantly, with stress being redistributed from the forefoot and rearfoot toward the midfoot. Previous studies also showed that arch-supporting insoles significantly reduced plantar fascia stress, with decreased peak stress at the rearfoot and forefoot accompanied by increased stress in the midfoot[12,28]. However, since kinesio tape provided relatively weaker support compared with insoles, this might have partly explained the relatively lower benefits observed.

Overall, both FO and kinesio tape demonstrated beneficial effects on lower limb kinematics, kinetics, and coordination, yet their mechanisms and magnitudes of influence differed. FO provided stronger structural support, which effectively corrected excessive eversion, limited dorsiflexion, and stabilized frontal-plane motion. These changes reduced coordination variability and improved in-phase knee–ankle coupling, thereby enhancing joint stability. However, the rigid support of FO increased knee adduction moment, which might have raised the risk of knee osteoarthritis in the long term. In contrast, kinesio tape, due to its elastic and less restrictive nature, exerted weaker corrective effects on ankle motion but contributed to running efficiency by lowering plantar propulsion forces and enhancing proprioceptive input. Its indirect support of the

transverse arch through neuromuscular facilitation promoted more effective muscle activation without imposing excessive mechanical constraints.

From a practical perspective, FOs might be more suitable for runners who require stronger correction and stability. Conversely, kinesio tape may be a preferable option for runners seeking a less invasive intervention to optimize running efficiency, reduce plantar load, and enhance muscle activation, particularly when long-term use or comfort during dynamic activities is prioritized.

4.3 FO impact on landing task under different heights

The findings showed that as landing height increased, the peak flexion angles and peak joint moments of the hip, knee, and ankle also increased. This pattern was consistent with prior work indicating that, under higher landing demands, the lower extremity tended to adopt larger flexion excursions and greater moment production to absorb more impact energy and dissipate mechanical load more effectively[155,156]. Biomechanically, greater flexion generally reflected a longer deceleration displacement and a more sufficient “cushioning range,” and together with higher joint moments, it distributed the impact over a longer absorption phase, thereby reducing the threat associated with abrupt peak loading.

In parallel, we observed that vertical ground reaction force (vGRF) and loading rate increased with landing height, which aligned with previous studies examining the effects of landing demands on impact characteristics[156-158]. Higher vGRF combined with a steeper loading rate indicated that impact accumulated more rapidly within a shorter time window, which potentially heightened the stress environment for articular cartilage, ligaments, and surrounding soft tissues, especially when landing alignment and neuromuscular control were suboptimal.

After the application of foot orthoses (FO), the ankle eversion angles at all three landing heights decreased significantly. This suggested that the FO restricted medial longitudinal arch collapse and excessive pronation during landing, thereby improving foot–ankle alignment and stability, which was also supported by previous evidence[77]. Because excessive pronation in flatfoot landing was often accompanied by a knee valgus

posture, which might increase knee joint stress, the improved foot alignment after FO use was accompanied by reduced frontal-plane knee valgus angles, demonstrating a trend similar to that reported in running-related literature[159]. In the transverse plane, arch support reduced excessive tibial internal rotation, and knee internal rotation amplitude also decreased, which might have lowered torsional stress at the knee at the instant of landing. Given that the internal-rotation chain associated with flatfoot could propagate proximally, hip posture also improved to some extent after FO intervention, as reflected by reduced hip adduction and hip internal rotation angles[159].

Moreover, the FO likely acted through a combination of arch support and material damping to prolong the force-application time and attenuate instantaneous impact, which slowed the rate at which vGRF reached its peak and reduced the loading rate. A lower loading rate implied a more gradual rise in impact, smaller instantaneous shock to the lower limb, and a potentially reduced risk of lower-extremity injury[160]. Notably, the effects of FO on peak valgus angles and moments at the knee and ankle were more evident at higher landing heights[158]. This trend implied that when landing impacts intensified, the protective role of FO became more important and the biomechanical benefits provided by insoles became more pronounced.

Across landing heights, the insole intervention consistently exerted positive effects, and its relative benefits became more apparent as landing impact increased. Therefore, FO could have served as a basic support in daily, low-intensity activities, while in high-intensity drop-landing scenarios it likely functioned as a more critical protective aid by buffering impact and optimizing lower-limb biomechanics.

4.4 Customized 3D-printed insoles: FEA and CNN

This study aimed to employ finite element methods (FEM) to simulate and compare the unloading capacity of 3D-printed insoles incorporating an auxetic structure and a “#”-shaped structure for diabetic feet. The primary objective was to mitigate the risk of foot ulcers arising from diabetes-related pathological changes in the foot, and an additional

attempt was made to use a convolutional neural network (CNN) algorithm to recommend optimal insole structures tailored to individuals with different body weights.

To achieve these aims, an assessment procedure for unloading insoles was devised and tested, and the influence of structural design on plantar pressure distribution in diabetic patients was examined in a systematic manner. Under comparable loading conditions, auxetic-structure insoles demonstrated more pronounced unloading effects than both barefoot conditions and non-auxetic structures, indicating that geometry-driven deformation mechanisms could meaningfully improve pressure redistribution.

Although negative Poisson's ratio structures were no longer considered novel, they had already shown potential applications in medical devices, sports equipment, sensors, aerospace engineering, and other fields[161]. Because negative Poisson's ratio structures exhibited a distinctive shape-adaptive deformation—characterised by overall lateral contraction of the material during uniaxial compression—they were regarded as having favourable prospects for insole applications. This potential was previously demonstrated by the negative Poisson's ratio heel pads designed and fabricated by Leung et al[162]. In that work, the negative Poisson's ratio structure underwent a unique deformation when the heel exerted pressure on the insole, which increased the heel–insole contact area and thereby reduced peak contact force. While such designs offered useful insights for diabetic foot protection, the cited modelling focused primarily on the heel region and did not extend to the full foot. Because walking generated high impact forces not only at the heel but also around the first and fifth metatarsophalangeal joints, it was therefore also important to analyse pressure responses across the whole foot rather than restricting attention to a single region.

Moreover, the study showed that pressure reduction effects were not solely determined by insole design; they also correlated with horizontal ground reaction force, which was plausibly related to body weight. This finding implied that, in practical clinical applications, it would likely be necessary to adjust structural parameters—such as the thickness of the insole lattice or unit-cell geometry—according to an individual's loading

conditions[163,164]. In particular, when a patient was overweight, a stiffer insole was expected to be required to better resist deformation and to maintain effective pressure offloading. A similar conclusion was reported by Chatzistergos et al.[163,165], who found a significant correlation between BMI and the optimal insole firmness; this provided supportive evidence for the trends observed in the present study. Nevertheless, although the optimally adapted firmness insole in Chatzistergos et al.'s work reduced pressure by an average of 46%—exceeding the 39.2% reported in this study—the difference could reasonably have been attributed to variations in structural design and material selection rather than a contradiction in the underlying biomechanical relationship.

Accordingly, it was suggested that future work could involve the selection of materials spanning different hardness levels and incorporating different density gradients, enabling more in-depth testing of how combined material–structure design influenced unloading performance. Such extensions were expected to clarify how stiffness tuning and graded architectures could be leveraged to improve plantar pressure redistribution across a wider range of patient body types and gait loading patterns.

In parallel, footwear design and manufacturing had undergone substantial advancement over the past decades with the emergence of data analytics, among which the application of machine learning had been particularly notable[166]. Conventional pipelines typically used kinematic variables, dynamic parameters, plantar pressures, and footwear characteristics as dataset inputs, and then trained machine learning algorithms to support classification, prediction, or design-selection tasks. For example, Malki et al. employed machine learning algorithms to aid the design of a rocker midsole intended to reduce plantar pressures, and their approach succeeded in controlling peak plantar pressure to within 200 kPa[167]. In the present study, the proposed methodology permitted the recommendation of an appropriately constructed insole based on body weight to a certain extent. However, the limitations of the amount of data and the ‘black box’ problem still pose a significant challenge to the dissemination and use of the technology[168].

4.5 Limitations

This study had several limitations. First, restricting the sample to female participants enhanced internal consistency and reduced variability attributable to sex differences, but it also limited the extrapolation of the findings to male or mixed-sex populations. To strengthen the stability of effect estimates and to examine potential sex-specific responses, the study design required larger and more diverse cohorts. Second, the study primarily focused on the acute biomechanical effects of foot orthoses and kinesio tape; therefore, it did not capture longer-term neuromuscular adaptations or possible “wash-out” effects associated with prolonged use. In practical applications, the mechanical efficacy of tape could also have been influenced by factors such as initial application tension and perspiration-related reductions in adhesive performance during activity, which might have led to subtle differences in sustained support of the longitudinal and transverse arches compared with the controlled laboratory setting. Third, the task protocol was restricted to standardized vertical drop-landing, and all measurements were collected under non-fatigued conditions; however, real-world sporting scenarios often involved multi-planar movements, and fatigue typically impaired neuromuscular control and joint stability. Consequently, it remained unclear whether the observed biomechanical improvements and impact-buffering effects would have persisted under more ecologically valid and fatigued conditions. Finally, technical and computational limitations related to data volume and modeling remained, and thus the current findings should have been interpreted primarily as a methodological proof-of-concept and reference framework rather than as quantitatively optimal, individualized prescriptive recommendations. In addition, the finite element analysis adopted simplified linear material properties, which might not have fully characterized the viscoelastic behavior of plantar soft tissues such as the plantar fascia; and due to data constraints, the machine learning workflow was not yet fully optimized. Future work was therefore needed to expand datasets through large-scale gait testing and to integrate more advanced multi-algorithm optimization frameworks to improve predictive accuracy and structural optimization.

5 CONCLUSIONS AND FUTURE WORKS

5.1 Conclusions

FO provided positive biomechanical effects for alleviating flatfoot symptoms but also caused biomechanical changes associated with increased loading at the knee and MTP joints in walking task. As hardness increased, changes in tibialis posterior peak muscle force and in ankle and foot biomechanics tended to plateau, and the beneficial effects were no longer obvious. At the MTP joint, however, greater hardness improved its biomechanical behavior while simultaneously increasing joint loading, and knee load-related parameters increased with hardness as well. Therefore, the selection of FO hardness should aim to achieve an optimal balance between maximizing corrective function and avoiding excessive increases in joint loading.

Both interventions provided significant biomechanical benefits, including reduced plantar fascia stress, corrected excessive ankle eversion, and improved knee–ankle coordination stability. Specifically, FOs provided stronger structural support that effectively stabilized frontal-plane motion and limited abnormal ankle displacement, although they also increased knee extension and adduction moments, which potentially raised the long-term risk of knee osteoarthritis. In contrast, kinesio tape exerted more subtle corrective effects but enhanced running efficiency and muscle activation through improved proprioceptive feedback and reduced plantar propulsion forces. Consequently, while FOs were found to be more suitable for runners requiring maximum structural correction and stability, kinesio tape offered a preferable, less invasive alternative for those prioritizing movement efficiency and comfort during running tasks.

While increased landing heights escalated joint flexion excursions and impact forces, the application of foot orthoses effectively mitigated these biomechanical risks by improving lower-limb alignment and attenuating impact. Specifically, the orthoses restricted excessive ankle eversion, knee valgus, and internal rotation across the kinetic chain, while simultaneously lowering the vertical ground reaction force loading rate through arch support and material damping. Furthermore, the protective benefits of the

intervention became more pronounced as landing demands increased, suggesting that foot orthoses served as a critical biomechanical aid for stabilizing the lower extremity and reducing potential injury risk during high-impact activities.

The feasibility and significance of applying personalized design and simulation methods to 3D-printed offloading insoles were verified. Under the same density and material hardness, the negative Poisson's ratio structure achieved better pressure relief by increasing the effective contact area with the skin. The offloading effect was associated with body weight, and patients with higher body weight might have obtained better pressure relief when using insoles with higher hardness. In addition, although the machine-learning classification performance in this study did not reach an ideal level, it still provided methodological value as a reference.

5.2 Future works

To strengthen the stability of effect estimation and to examine potential sex-specific responses, future research needed to expand both sample size and population heterogeneity by including broader participant profiles (e.g., males, different age strata, different body mass index (BMI)/body weight, varying severity of pes planus, and diverse foot-shape parameters). In addition, the existing evidence primarily captured immediate biomechanical changes under acute testing conditions, whereas foot orthoses in real-world sport and daily use were typically accompanied by prolonged wear, material property degradation, neuromuscular adaptation, and cumulative fatigue. Accordingly, subsequent work also needed to systematically evaluate the long-term effects of the two interventions and their time-dependent trajectories.

Building on an integrated perspective that combined finite element modeling (FEM) with machine learning, a dedicated “data–simulation–learning–manufacturing–validation” workflow was required for the design of flatfoot orthotic insoles to enable scalable, individualized prescriptions. This workflow was envisaged to include standardized data acquisition by integrating gait kinematics/kinetics, plantar pressure, 3D foot scans or image-based geometry, body weight/BMI, arch height indices, and hindfoot alignment,

while unifying walking/running speed, footwear type, and task protocols to reduce noise; individualized FEM construction and parameter calibration incorporating foot bones, soft tissues, and plantar fascia with progressively more realistic material behaviors (nonlinearity, viscoelasticity, and frictional contact) and calibration against measured plantar pressure/deformation to improve credibility; multi-objective optimization that simultaneously targeted corrective efficacy and safety constraints (e.g., mitigating excessive eversion/pronation, reducing plantar fascia stress or tibialis posterior demand, and imposing upper bounds on knee adduction moment, knee joint reaction forces, and MTP joint peak reaction force to prevent proximal or distal load transfer); machine-learning-based recommendation of structural parameters (stiffness, zonal stiffness distribution, lattice type/density, and arch-support height) with interpretability and uncertainty quantification to support clinical usability; and rapid manufacturing via 3D printing followed by closed-loop validation through in-shoe pressure, gait metrics, and perceived comfort across multiple visits or training cycles. Through this workflow, flatfoot orthosis design was expected to shift from experience-driven tuning toward evidence-driven individualized optimization, prioritizing not only correction magnitude but also long-term safety under sustained wear and athletic applications.

NEW SCIENTIFIC THESIS POINTS

1st Thesis point

I applied a special method: Three-dimensional motion capture, force-plate, and EMG data were integrated with OpenSim musculoskeletal simulations, and Bayesian repeated-measures ANOVA together with Bayesian correlation analyses were used to evaluate changes in joint ROM, peak moments, joint reaction forces, and muscle forces.

During walking the orthoses generally improved rearfoot control by reducing coronal-plane ROM, peak moments, and joint reaction forces at the ankle, subtalar, and midtarsal joints, while decreasing tibialis posterior peak muscle force; however, these rearfoot gains plateaued rather than increasing linearly with hardness. In contrast, greater hardness progressively shifted loading to other regions: under the hardest condition, knee coronal ROM and moment increased, knee sagittal peak joint reaction force and vastus medialis peak muscle force rose, and forefoot demand intensified, as reflected by higher metatarsophalangeal ROM, moment, and joint reaction force.

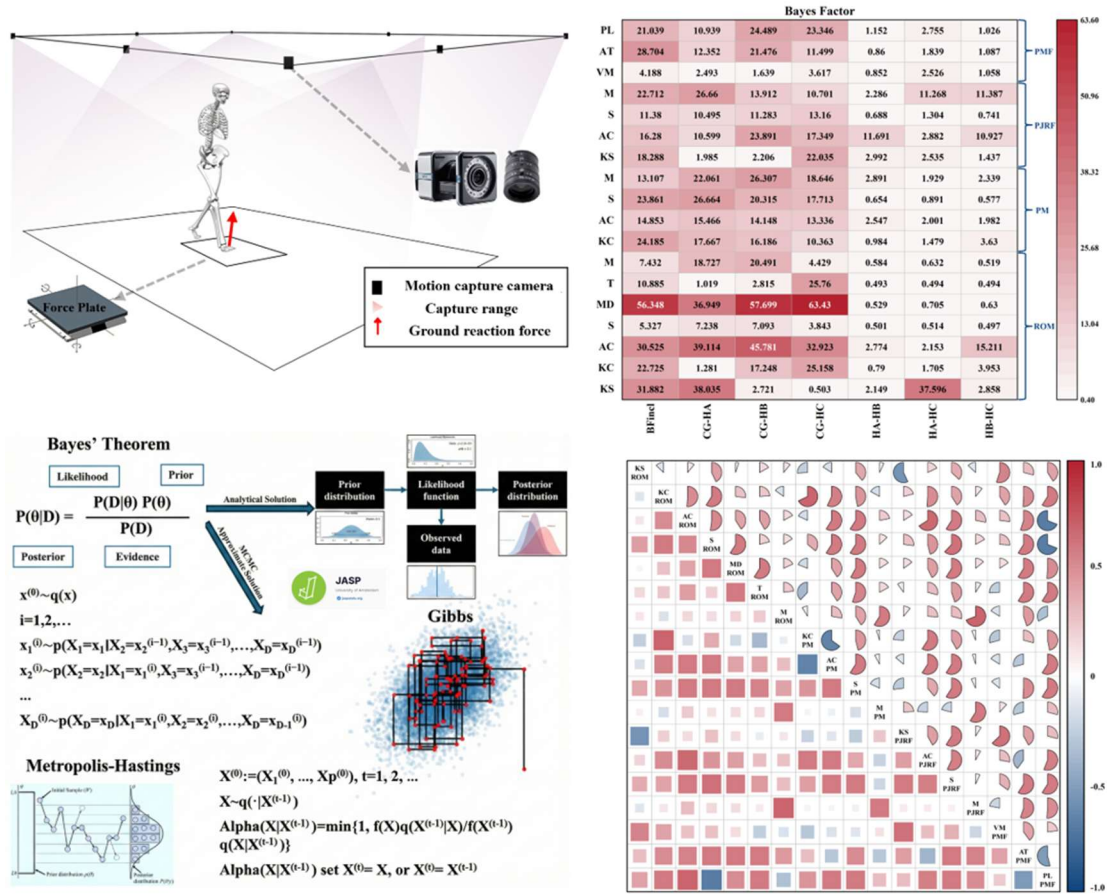


Figure 1 BF₁₀ values under different arch hardness conditions (up) and correlation coefficient between different biomechanical parameters (down)

Related articles to the first thesis point:

1. **Chen H**, Sun D, Zhang Q, et al. Biomechanical Effects of Varying Arch Support Hardness in Foot Orthosis for Adults with Flexible Flatfoot: A Comprehensive Bayesian Statistical Analysis[J]. Gait & Posture, 2025: 110085.
2. **Chen H**, Sun D, Fang Y, et al. Effect of orthopedic insoles on lower limb motion kinematics and kinetics in adults with flat foot: a systematic review[J]. Frontiers in Bioengineering and Biotechnology, 2024, 12: 1435554.

2nd Thesis point

I examined flexible flatfoot runners and compared the multi-level biomechanical effects of foot orthoses (FOG) versus kinesio tape (KTG) during running.

I found that the applied interventions improved running biomechanics in flexible flatfoot, yet their mechanisms and magnitudes differed in measurable ways. Both FOG and KTG reduced excessive pronation-related behavior, reflected by lower peak ankle eversion and enhanced knee–ankle coordination stability. However, FOG provided stronger mechanical restraint, further limiting dorsiflexion/eversion and reducing plantar fascia as well as forefoot and rearfoot stresses, while simultaneously shifting loading demands proximally, as shown by increased knee adduction and extension moments that potentially implied long-term joint-loading concerns. In contrast, KTG exerted milder kinematic correction but modified propulsion dynamics, with lower anteroposterior GRF signal power and delayed maximum power timing, consistent with enhanced proprioceptive control and potentially improved running efficiency.

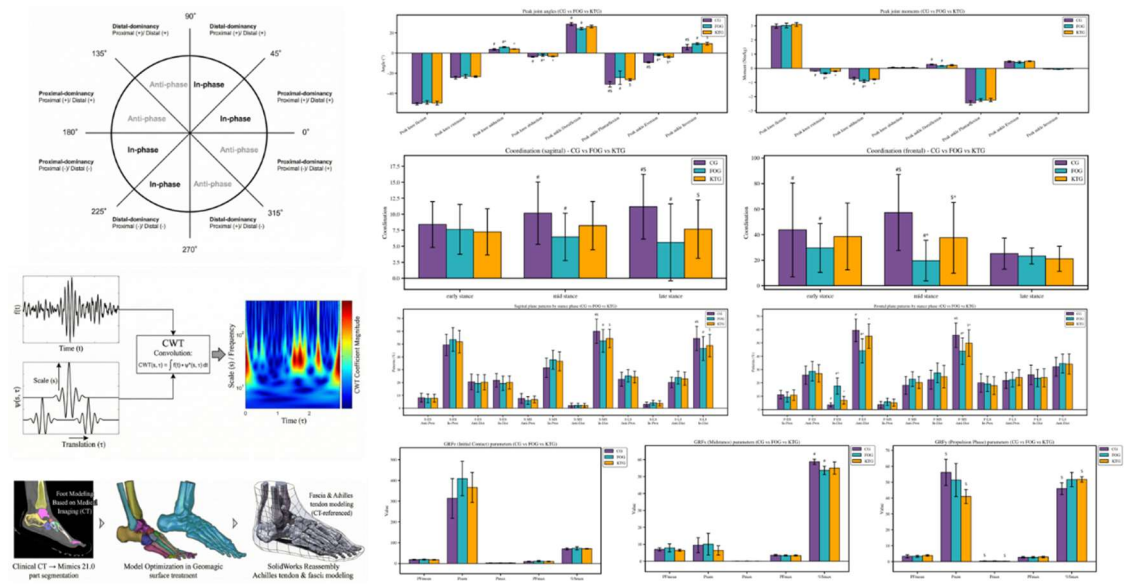


Figure 34 Peak angles and moments, knee-ankle coordination (variability and patterns), and time-frequency GRF under foot orthoses and kinesio tape conditions

Related articles to the second thesis point:

1. **Chen H**, Bíró I. Effects of Foot Orthoses and Kinesio Tape on Spatiotemporal and Kinetic Gait Parameters during Running in Individuals with Flatfoot[J]. *Analecta Technica Szegedinensia*, 2024, 18(3): 77-81.
2. **Chen H**, Sun D, Zhang Q, et al. Biomechanical Differences Between Habitual Forefoot and Rearfoot Strike Running: A Systematic Review[J]. *Advanced Exercise and Health Science*, 2024.

3rd Thesis point

I evaluated how custom-made arch-support orthoses influenced landing biomechanics in individuals with flatfoot across increasing drop heights (20, 40, 60 cm). Flatfoot participants performed controlled landings with and without custom orthoses, while lower-limb kinematics, joint moments, vertical ground reaction forces (vGRF), and loading rates (LR) were recorded. As landing height increased, both conditions showed larger peak hip/knee/ankle flexion angles and moments, along with higher vGRF and LR, indicating greater shock-absorption demands at higher drops. Compared with no-insole landings, the custom orthoses consistently reduced ankle eversion at all heights, reflecting constrained pronation-related mechanics. They also improved proximal alignment within the kinetic chain, yielding smaller hip adduction/internal rotation, knee abduction (valgus), and knee internal rotation angles. In kinetics, orthoses lowered peak knee abduction and ankle eversion moments and reduced forefoot and rearfoot loading rates, suggesting a slower, less abrupt rise in impact forces. The percentage reductions in valgus-related angles and moments were greatest at 60 cm, implying that orthotic benefits became more evident as impact intensity increased.

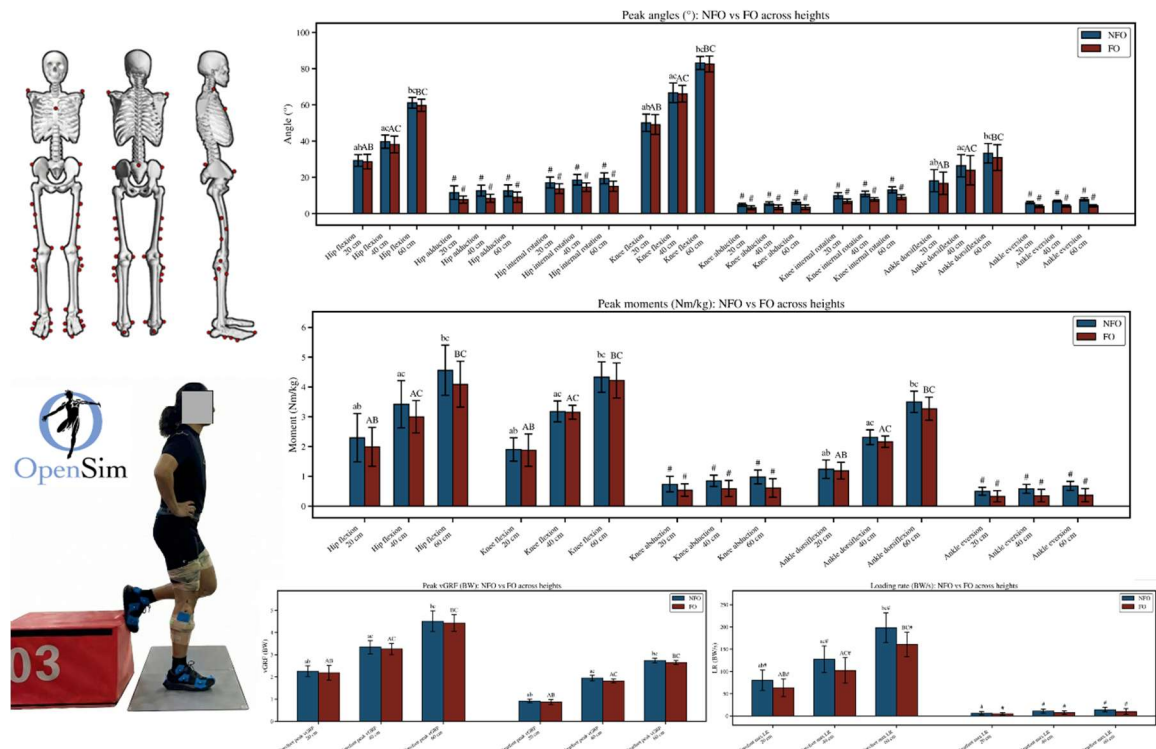


Figure 35 Peak joint angles, moments, and ground reaction forces under 20, 40, 60 cm landing heights with and without FO

Related articles to the third thesis point:

1. **Chen H**, Bíró I. Advances in the application of orthotic insoles for flatfoot treatment [C]. Proceedings of the Engineering Symposium at Bánki, 2024, pp: 42-49.
2. **Chen H**, Zhang Q, Bíró I, et al. Effects of Different Longitudinal Bending Stiffness Shoes on Distal Joint Kinematics and Muscular Mechanics in Adolescent Amateur Runners[M]. Material Strength and Applied Mechanics. IOS Press, 2024: 472-478.

4th Thesis point

I developed and evaluated personalized 3D-printed insoles for diabetic foot care by integrating finite element analysis (FEA) with an exploratory machine-learning workflow. A subject-specific foot–insole model was reconstructed from CT imaging, and plantar pressure plate measurements were used to identify high-pressure regions and guide zoned insole design for targeted offloading. Two lattice types were compared: an auxetic (negative Poisson’s ratio) structure and a conventional structure, with regional thickness gradients (1–3 mm) implemented in high-load areas. FEA simulations constrained the proximal foot and applied vertical loads (750–1350 N) to represent different loading/weight conditions, and simulated barefoot pressure patterns were compared against experimental pressure distributions to support model credibility.

I have combined CT and pressure-guided customization with FEA-based parametric optimization and a CNN-based recommendation attempt, thereby providing a transferable methodological framework for scalable personalized insole design while demonstrating that auxetic structures had delivered superior, load-dependent plantar-pressure offloading.

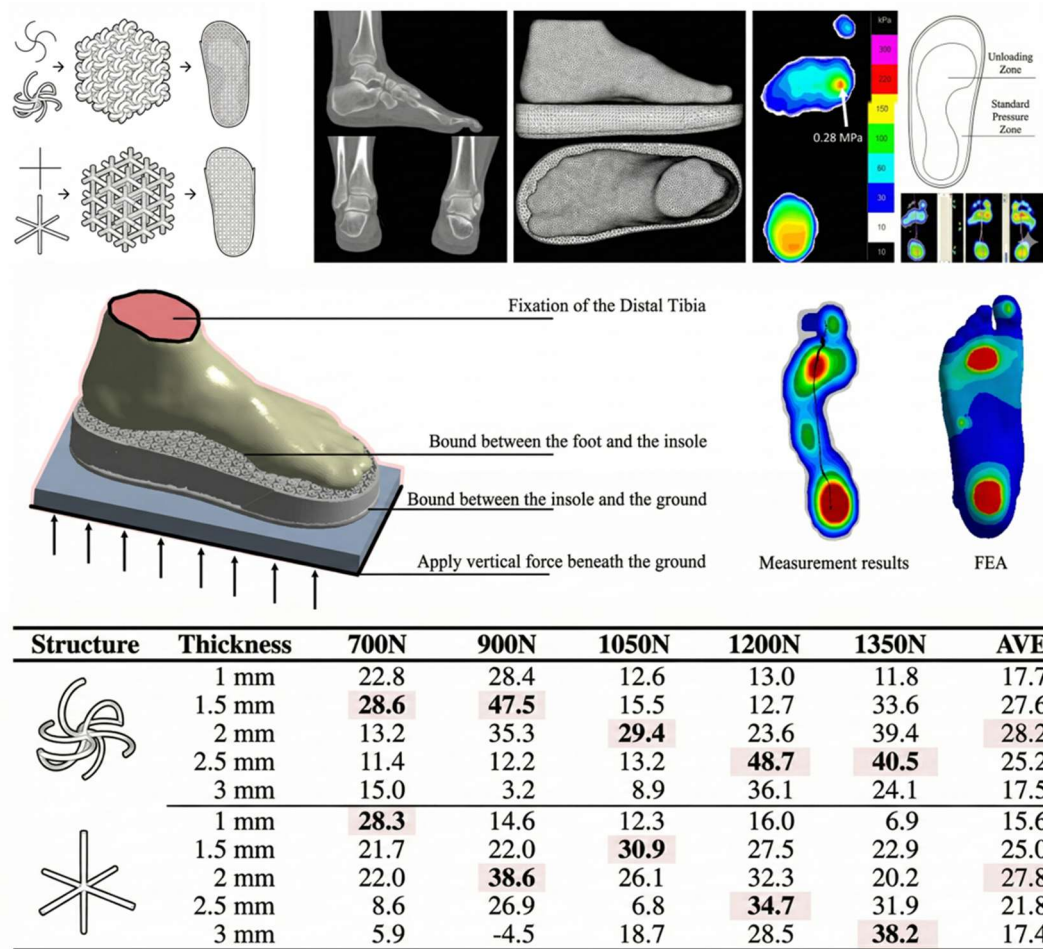


Figure 36 Pressure redistribution and reduction of auxetic structure and “#” shaped structure insoles of varying thicknesses under different body weight simulations

Related articles to the fourth thesis point:

1. Lu Z, Li X, Gao Z, **Chen H**, et al. Customized 3D-Printed Insoles for Diabetic Foot Care: Finite Element Analysis and Machine Learning Approach[M]. Material Strength and Applied Mechanics. IOS Press, 2024: 515-522.
2. Zhang Q, Sun D, Wang M, Tafferner-Gulyás V, **Chen H**, et al. Effect of plantar fascia stiffness on plantar windlass mechanism and arch: Finite element method and dual fluoroscopic imaging system verification[J]. Medical Engineering & Physics, 2024, 134: 104259.

LIST OF PUBLICATIONS

Articles related to this thesis

1. **Chen H**, Sun D, Fang Y, et al. Effect of orthopedic insoles on lower limb motion kinematics and kinetics in adults with flat foot: a systematic review[J]. *Frontiers in Bioengineering and Biotechnology*, 2024, 12: 1435554. **IF:4.8, JCR: Q1**
2. **Chen H**, Sun D, Zhang Q, et al. Biomechanical Effects of Varying Arch Support Hardness in Foot Orthosis for Adults with Flexible Flatfoot: A Comprehensive Bayesian Statistical Analysis[J]. *Gait & Posture*, 2025: 110085. **IF:2.4, JCR: Q2**
3. **Chen H**, Zhang Q, Bíró I, et al. Effects of Different Longitudinal Bending Stiffness Shoes on Distal Joint Kinematics and Muscular Mechanics in Adolescent Amateur Runners[M]. *Material Strength and Applied Mechanics*. IOS Press, 2024: 472-478. **EI**
4. **Chen H**, Bíró I. Effects of Foot Orthoses and Kinesio Tape on Spatiotemporal and Kinetic Gait Parameters during Running in Individuals with Flatfoot[J]. *Analecta Technica Szegedinensia*, 2024, 18(3): 77-81.
5. **Chen H**, Sun D, Zhang Q, et al. Biomechanical Differences Between Habitual Forefoot and Rearfoot Strike Running: A Systematic Review[J]. *Advanced Exercise and Health Science*, 2024, 1(4): 241-247.
6. **Chen H**, Bíró I. Advances in the application of orthotic insoles for flatfoot treatment [C]. *Proceedings of the Engineering Symposium at Bánki*, 2024, pp: 42-49.
7. Lu Z, Li X, Gao Z, **Chen H**, et al. Customized 3D-Printed Insoles for Diabetic Foot Care: Finite Element Analysis and Machine Learning Approach[M]. *Material Strength and Applied Mechanics*. IOS Press, 2024: 515-522. **EI**
8. Zhang Q, Sun D, Wang M, Tafferner-Gulyás V, **Chen H**, et al. Effect of plantar fascia stiffness on plantar windlass mechanism and arch: Finite element method and dual fluoroscopic imaging system verification[J]. *Medical Engineering & Physics*, 2024, 134: 104259. **IF:2.3, JCR: Q3**

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1. Pan J, **Chen H**, Zheng Z, et al. A comparative analysis of bionic and neutral shoes: impact on lower limb kinematics and kinetics during varied-speed running[J]. Applied Sciences, 2023, 13(23): 12582. **IF:2.5, JCR: Q2**
2. Lu R, **Chen H**, Huang J, et al. Biomechanical investigation of lower limbs during slope transformation running with different longitudinal bending stiffness shoes[J]. Sensors, 2024, 24(12): 3902. **IF:3.5, JCR: Q2**
3. Cai J, Sun D, Xu Y, **Chen H**, et al. The influence of medial and lateral forefoot height discrepancy on lower limb biomechanical characteristics during the stance phase of running[J]. Applied Sciences, 2024, 14(13): 5807. **IF:2.5, JCR: Q2**
4. Zhu A, Gao S, Huang L, **Chen H**, et al. Effects of fatigue and unanticipated factors on knee joint biomechanics in female basketball players during cutting[J]. Sensors, 2024, 24(14): 4759. **IF:3.5, JCR: Q2**
5. Li F, Sun D, Zhang Q, **Chen H**, et al. Effect of a Double Helical Spring Decompression Structure Backpack on the Lumbar Spine Biomechanics of School-Age Children: A Finite Element Study[J]. Molecular & Cellular Biomechanics, 2023, 20(1). **EI**
6. Zhang Q, Sun D, Song Y, **Chen H**, et al. Effects of Plantar Fascia Stiffness on Windlass Mechanism: A Finite Element Analysis[J]. Journal of medical biomechanics, 2025, 40(1): 93-99.
7. Gao S, Sun D, Song Y, Cen X, **Chen H**, et al. Induced fatigue impact on plantar pressure in females with mild hallux valgus[J]. Molecular & Cellular Biomechanics, 2024, 21: 135-135. **EI**
8. Gao S, Song Y, Sun D, Zheng Z, **Chen H**, et al. The impact of running experience and shoe longitudinal bending stiffness on lower extremity biomechanics: a cross-sectional study[J]. Acta of Bioengineering and Biomechanics, 2025, 26(2): 93-103. **IF:0.8, JCR: Q4**

9. Hu C, Song Y, Sun D, Lu Z, **Chen H**, et al. Sedentary Duration and Systemic Health Burden: Nonlinear Associations with Muscle, Fat, and Vascular Phenotypes in a US Population-Based Study[J]. Healthcare, 2025, 13(18): 2309. **IF:2.7, JCR: Q2**
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12. **Chen H**, Zhang Q, Biró I. Effects of Arch Support Pads and Insoles on Gait Parameters and Plantar Mechanics During Running in Adults with Flatfoot[C]. 2024 IEEE 7th International Conference and Workshop Óbuda on Electrical and Power Engineering (CANDO-EPE). IEEE, 2024: 99-102.
13. **Chen H**, Zhang Q, Biró I. A Comparative Analysis of Different Longitudinal Bending Stiffness Shoes on Running Biomechanics in Adolescent Amateur Runners[C]. 2024 IEEE 18th International Symposium on Applied Computational Intelligence and Informatics (SACI). IEEE, 2024: 83-88.
14. Zhang Q, **Chen H**, Biró I, et al. Biomechanical Characteristics of Adolescent Cervical Forward Flexion Analyses Based on the Finite Element Method[M]. Material Strength and Applied Mechanics. IOS Press, 2024: 479-485. **EI**
15. Zhang Q, **Chen H**, Biró I. Effect of stiffness difference of forefoot transverse arch on foot biomechanics[C]. 2024 IEEE 7th International Conference and Workshop Óbuda on Electrical and Power Engineering (CANDO-EPE). IEEE, 2024: 53-58.
16. Zhang Q, **Chen H**, Jiang X, et al. Prediction and Effectiveness Verification of Chevron Osteotomy Surgical Plan Using Finite Element Method[C]. 2024 IEEE 18th International Symposium on Applied Computational Intelligence and Informatics (SACI). IEEE, 2024: 89-94.

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ABBREVIATION

AAFD: Adult-acquired flatfoot deformity

AFOs: ankle-foot orthoses

AM: Additive manufacturing

ANOVA: analysis of variance

AT: anterior tibialis

ATL: achilles tendon loading

ATLR: achilles tendon loading rate

BMI: body mass index

BT: big toe

CASO: custom arch support orthotics

CG: control group

CMC: computed muscle control

CMJ: countermovement jump

CNN: Convolutional Neural Network

CNS: central nervous system

COP: center of pressure

CWT: Continuous wavelet transforms

EMG: electromyography

EVA: ethylene-vinyl acetate

FE: finite element

FEA: finite element analysis

FEM: finite element modeling

FFF: Fused filament fabrication

FO: foot orthosis

FOG: foot orthosis group

FOs: foot orthoses

GRF: ground reaction force

GRFs: ground reaction forces

HL: Heel Lifts

ID: inverse dynamics

IK: inverse kinematics

KTG: kinesio tape group

LDT: Low-Dye Taping

LG: lateral gastrocnemius

LH: lateral heel

M2, M3, M4: second, third, and fourth metatarsals

MG: medial gastrocnemius

MH: mid-heel

MLA: medial longitudinal arch

MTP: metatarsophalangeal

MTP2,3,4,5: second, third, fourth, and fifth metatarsophalangeal

MWFOs: Modified wedge foot orthoses

PFmax: pseudo-frequency

PFmean: mean pseudo-frequency

PFPS: patellofemoral pain syndrome

PL: peroneus longus

Pmax: maximum power

PPRI: plantar pressure redistribution insoles

Psum: sum of the signal power

PTT: posterior tibialis tendon

PTTD: posterior tibial tendon dysfunction

RRA: residual reduction algorithm

SBJ: standing broad jump

SLS: Selective laser sintering

SOL: soleus

STJ: subtalar joint

T: toe region

TFOs: toe-only footwear

TFOs: total contact foot orthoses

TP: tibialis posterior

TPU: thermoplastic urethane

vGRF: vertical ground reaction force

VL: vastus lateralis

VM: vastus medialis

%Smax: maximum power percent of stance

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