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An innovative biomechanical
exploration of dual Dual-plane
Fluoroscopy imaging for finite
element modelling of the lower limb

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DECLARATION AGAINST PLAGIARISM AND CORRECT REFERENCING

I, Qiaolin Zhang, declare that the dissertation entitled: An innovative biomechanical exploration of dual Dual-plane Fluoroscopy imaging for finite element modelling of the lower limb is my original work where the references given in the Reference list are used. All sections, which are transcribed or rewritten from some publications, they are correctly referenced.

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(Signature)

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Motivation of the work

The structure and function of human lower limb joints are the products of long-term evolutionary adaptation, profoundly reflecting the biomechanical evolution from arboreal life to terrestrial bipedalism. Compared with non-human primates and archaic humans, modern humans have evolved a larger intercondylar notch and broader joint surfaces in the knee to provide superior load-bearing capacity and movement stability [1]. Regarding the ankle joint, while apes prioritize flexibility for climbing, humans have evolved a tibiotalar joint with a larger contact area and a more vertical orientation. This adaptation significantly enhances ankle stability to meet the weight-bearing demands of upright walking [2]. This biomechanical characteristic of "structure determines function" implies that distinct joint morphologies correspond to specific movement patterns. Therefore, an in-depth investigation into the kinematic characteristics of the human knee, ankle, and foot is of significant scientific importance for understanding human evolution and elucidating the biomechanical mechanisms of modern human joints.

However, with the intensification of global population aging and the widespread popularity of national fitness activities, functional impairments, degenerative diseases, and sports injuries related to lower limb joints have become increasingly frequent, emerging as urgent social issues [3]. For instance, knee osteoarthritis has become the leading cause of mobility limitation in the elderly, while Chronic Ankle Instability often leads to recurrent sprains and long-term pain [4-6]. Since abnormal joint kinematics are often regarded as key factors inducing the progression of osteoarthritis and lower limb overuse injuries, precise kinematic analysis is crucial for disease prediction and early intervention. Furthermore, the development of intelligent medical devices, rehabilitation exoskeletons, and bionic robots has placed higher demands on the accuracy of lower limb joint motion data [7, 8]. Consequently, it is particularly urgent to deeply explore the motion laws of human joints, especially the knee, ankle, and foot, under dynamic loading.

Although traditional kinematic analysis methods, such as skin-marker-based optical capture systems and multi-segment models, have been widely applied [6], their core

defect lies in the Soft Tissue Artifact (STA). This phenomenon refers to the relative sliding between markers on the skin surface and the underlying bone, making it difficult to precisely reproduce the true in vivo movement of joints. In terms of medical imaging, X-rays can only provide two-dimensional projections, while Computed Tomography (CT) and Magnetic Resonance Imaging (MRI), despite their ability to reconstruct three-dimensional anatomical structures, are limited to static scanning in a supine position. Thus, they cannot reflect physiological characteristics under weight-bearing or dynamic conditions [9]. To overcome this bottleneck, the Dual-plane Fluoroscopy Imaging System (DFIS) was developed. By combining high-frame-rate dynamic X-ray images with high-resolution anatomical models from CT or MRI, this technology achieves real-time, six-degrees-of-freedom (6-DOF) precise capture of skeletal motion. This capability makes it possible to analyze complex human movements with sub-millimeter accuracy, providing revolutionary technical support for biomechanical research [10].

Beyond acquiring precise kinematic data, further exploring the internal mechanical environment of the joint represents another major challenge in biomechanics. The Finite Element Method (FEM), as a powerful computational simulation tool, has been widely used to assess the internal biomechanical environment of complex structures like the foot and ankle, particularly the effects of loading on bones and soft tissues [11]. However, the predictive accuracy of Finite Element (FE) models is highly dependent on the accuracy of boundary conditions. Traditional FEA often adopts idealized or generic motion boundary conditions, which struggle to reflect the individual's authentic motion state. This is where the advantage of DFIS becomes prominent: it can provide high-precision, subject-specific dynamic boundary conditions for FE models. By using the real skeletal kinematic data captured by DFIS to directly drive the FE model, it is possible to accurately simulate the stress and strain distributions within bones, cartilage, and soft tissues during real running or movement processes [12]. This coupling method of "experimental measurement and computational simulation" effectively compensates for the dual limitations of traditional motion analysis and traditional FEA.

This study aims to establish an innovative biomechanical analysis framework that deeply integrates Dual-plane Fluoroscopy technology with Finite Element Analysis (FEA). By introducing high-precision in vivo motion data as driving conditions into the computational model, this research expects to reveal the intrinsic mechanical mechanisms of the lower limb under dynamic loads. This approach aims to significantly improve the biological fidelity and predictive accuracy of FE models, thereby providing a robust scientific basis for investigating the mechanisms of running injuries, formulating rehabilitation strategies, and optimizing sports performance.

Research objectives

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

The first research objective: To develop and analyze a foot biomechanical model using a multimodal data fusion approach. Initially, a detailed three-dimensional FE model comprising 29 bones, ligaments, and soft tissues was constructed based on CT imaging data from the subject. To establish precise load boundary conditions for this model, muscle forces for key groups such as the gastrocnemius and soleus were calculated using an OpenSim musculoskeletal model, driven by gait data acquired from a Vicon motion capture system and force plates. Model validation was conducted by comparing experimental measurements with simulated values for the medial longitudinal arch (MLA) length, height, and the rearfoot transverse arch width, utilizing the DFIS to capture in vivo skeletal motion during walking. Subsequently, the study simulated the Windlass mechanism under a condition of 30 degrees of dorsiflexion at the first metatarsophalangeal joint. By varying the Young's modulus of the plantar fascia from 25% to 200% of the reference value (87.5 MPa to 700 MPa), the system systematically quantified the specific effects of different fascia stiffness levels on the morphology of the MLA, the dynamics of the rearfoot transverse arch, the metatarsophalangeal joint dorsiflexion angle, and the internal stress distribution within the plantar fascia.

The primary aim of this research is to establish an innovative framework combining dual-plane fluoroscopy and FEA to investigate the regulatory mechanisms of plantar fascia stiffness on the Windlass mechanism and foot arch function. The core objective is to overcome the limitations of traditional motion capture in quantifying internal skeletal micro-motion by using DFIS technology to validate the effectiveness of the FE model, thereby obtaining a high-precision predictive model for foot mechanics. Building on this, the study specifically addresses the gap in existing literature regarding the impact of the Windlass mechanism on the rearfoot transverse arch, clarifying how plantar fascia stiffness synergistically influences the stiffness and morphological changes of both the

medial longitudinal and rearfoot transverse arches. Furthermore, by revealing the distribution pattern where increased fascia stiffness leads to stress concentration in the middle and proximal sections of the fascia, this research expects to elucidate the underlying mechanical mechanisms of injuries such as plantar fasciitis. These findings will provide a scientific theoretical basis for understanding human foot energetics, preventing and treating foot-related diseases, and designing assistive devices like orthotic insoles.

The second research objective: To constructed an individualized multimodal analysis framework integrating medical imaging and engineering simulation. The experimental design included five distinct step heights ranging from 120 mm to 200 mm. The DFIS was employed to capture in vivo dynamic X-ray images of the knee joint at a frequency of 60 Hz during stair climbing, while CT scans were used to reconstruct three-dimensional anatomical models of the tibia, femur, fibula, and patella. Simultaneously, kinematic and kinetic data were collected using a Vicon motion capture system, portable force plates, and a Delsys electromyography system. These data were imported into OpenSim to calculate muscle forces for the quadriceps, hamstrings, and gastrocnemius. Based on these inputs, a quasi-dynamic FE model of the tibiofemoral joint was established to simulate key moments of the single-leg stance phase. This model quantitatively calculated and analyzed the stress distribution and variation patterns in the medial and lateral menisci, as well as the anterior and posterior cruciate ligaments, across different step heights.

The principal objective of this study is to resolve the limitations of traditional three-dimensional kinematic analysis, which cannot capture internal joint load transmission, and static FE models, which neglect real motion trajectories and muscle coordination. By integrating optimized dual-plane fluoroscopy registration techniques, individualized FE modeling, and musculoskeletal simulation, the research aims to achieve a realistic assessment of the internal stress state of the knee during stair climbing. The core goal is to explore how step height, a critical geometric factor, influences the biomechanical response of the knee menisci and cruciate ligaments, identifying specific stress

concentration areas and growth trends as step height increases. Ultimately, this research expects to reveal potential mechanical factors leading to degenerative joint diseases, thereby providing theoretical grounds and data support for preventing knee injuries, optimizing barrier-free stair design standards, and formulating targeted rehabilitation training strategies.

The third research objective: To establish a comprehensive biomechanical analysis framework fusing musculoskeletal modeling, FEA, and a high-speed DFIS. First, high-precision anatomical data of the knee skeletal and soft tissue structures were obtained via CT and MRI scans. The in vivo skeletal motion trajectories during the landing phase of a stop-jump task were precisely captured using the DFIS system at 30 Hz, while kinematic and kinetic parameters were collected via Vicon motion capture, force plates, and electromyography signals. Based on this dataset, individualized knee FE models were constructed for six key instants during the landing process. Joint moments and quadriceps forces calculated via OpenSim were applied as precise boundary conditions and load inputs for transient analysis. Additionally, the study employed ridge regression analysis to systematically explore the quantitative relationship between three-dimensional knee moments, quadriceps force, and the maximum equivalent stress on the patellar tendon (PT). The model was rigorously validated using both electromyography signals and anterior tibial translation experiments.

This research aims to address the limitations of previous studies that largely relied on static or simplified models, which struggle to authentically reflect complex joint mechanical behaviors under high-impact movements. It is dedicated to revealing the in vivo biomechanical behavior of the PT during the stop-jump landing phase. The core purpose is to quantify the stress distribution characteristics of the PT during landing, identify the timing and specific regions of maximum stress, such as the bone attachment sites, and determine the key biomechanical factors governing PT loading, including knee flexion moments and quadriceps force. By establishing a predictive model connecting load to stress, this study expects to provide deep insights into the etiology of patellar tendinopathy, often known as jumper's knee. It also aims to offer a scientific basis for

preventing sports injuries, formulating rehabilitation strategies, and developing future real-time injury risk prediction systems based on machine learning.

1. LITERATURE REVIEW

1.1 Kinematic and Dynamic Testing Methods in Biomechanics

Human motion is the result of precise coordination between the skeletal, muscular, ligamentous, and nervous systems, exhibiting high complexity and multi-system coupling characteristics [13]. Biomechanics, as an interdisciplinary field integrating anatomy, physiology, and mechanical principles, aims to qualitatively or quantitatively analyze mechanical phenomena within living organisms [14]. Depending on the research subject, biomechanics can be categorized into biological solid mechanics, biological fluid mechanics, and sports biomechanics. Regarding the lower limb musculoskeletal system, research primarily focuses on Sports Biomechanics and Orthopaedic Biomechanics. These fields utilize theories such as multi-body dynamics and continuum mechanics to analyze the mechanical mechanisms of the human body during physiological activities, injury occurrence, and rehabilitation processes [15]. Their core significance lies in providing a scientific basis for clinical diagnosis, injury prevention, rehabilitation engineering, and implant design.

Biomechanical research relies heavily on the precise measurement and modeling of the kinematics and dynamics of human motion [16]. Kinematics focuses on spatiotemporal changes in position, velocity, acceleration, and joint angles, while dynamics concentrates on the transmission of forces, moments, power, and energy. The two are mutually complementary; relying solely on kinematics makes it difficult to explain load distribution within tissues, whereas relying solely on dynamics makes it hard to pinpoint the structural causes of specific movement patterns.

Currently, Optical Motion Capture systems based on optoelectronic reflective markers and multi-camera triangulation serve as the "gold standard" in gait analysis laboratories, as shown in **Figure 1**. This technology offers advantages such as high sampling rates, high spatial resolution, non-invasiveness, and good ecological validity. However, its core challenge lies in the STA. This phenomenon involves relative sliding between markers on the skin surface and the underlying bone, making it difficult to accurately capture true

skeletal motion. This error is particularly significant in areas with thick soft tissue coverage or complex joint degrees of freedom, such as the foot, ankle, and shoulder [6, 17, 18].

To address this limitation, early studies often employed robotic systems to drive cadaveric specimens for simulation testing [19]. Although this allows for direct measurement of bone motion, in vitro experiments cannot reproduce active muscle contraction or neural control, failing to reflect the true physiological state. Another method involves the invasive implantation of bone pins connected to external markers [20]. While this achieves extremely high precision, its invasive and traumatic nature limits its application to a very small number of specific studies, preventing its generalization in clinical or large-sample research. Therefore, there is an urgent need for a high-precision measurement method that is capable of in vivo assessment while effectively eliminating soft tissue artifacts.

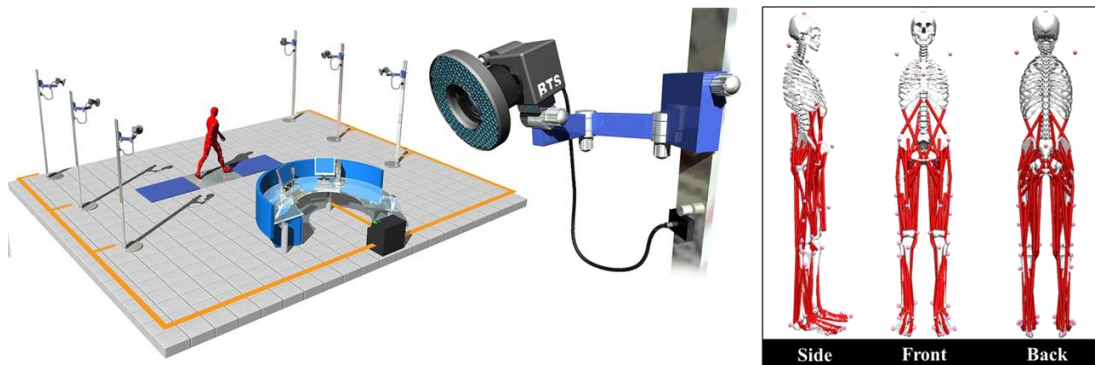


Figure 1. Optical motion capture system based on optoelectronic reflective markers and multiple cameras

1.1.1 Dual-plane Dynamic Fluoroscopy Imaging Technology

Traditional medical imaging techniques play a pivotal role in the assessment of skeletal morphology. X-ray radiography provides high-resolution two-dimensional projections, whereas CT and MRI enable the reconstruction of detailed three-dimensional anatomical structures, as illustrated in **Figure 2**. However, these modalities are predominantly confined to static or quasi-static imaging in a supine position. Consequently, they fail to capture joint behavior under weight-bearing or dynamic conditions, presenting a

significant limitation in elucidating the complex pathomechanical mechanisms of movement [21].

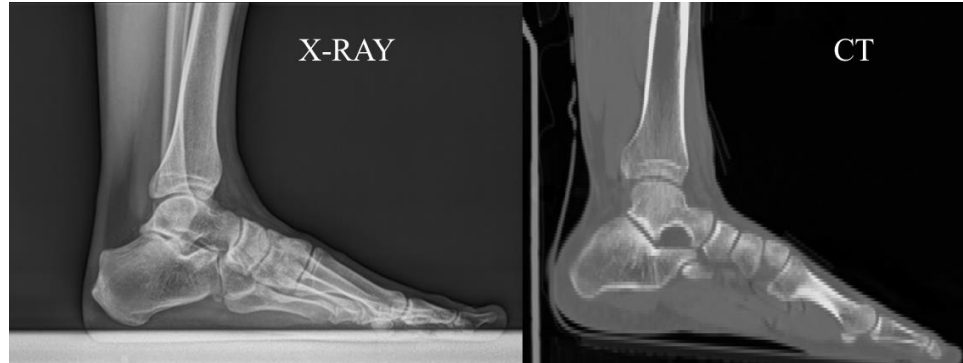


Figure 2. X-ray and CT images of the foot

In recent years, the DFIS also referred to as biplanar videoradiography, has emerged as a solution to the challenges [9]. This system primarily consists of two sets of angularly positioned X-ray emission and reception units, comprising high-voltage generators, X-ray tubes, image intensifiers, and high-speed cameras, alongside control and analysis software. Its operating principle involves the synchronous acquisition of two-dimensional fluoroscopic sequences of joint motion via two high-speed cameras. By employing 2D-3D registration algorithms, the system precisely aligns the subject's high-resolution static three-dimensional model, derived from CT or MRI, with the dynamic two-dimensional fluoroscopic images in a virtual space, as shown in **Figure 3** [22]. This process facilitates the reconstruction of continuous 6-DOF skeletal motion within three-dimensional space.

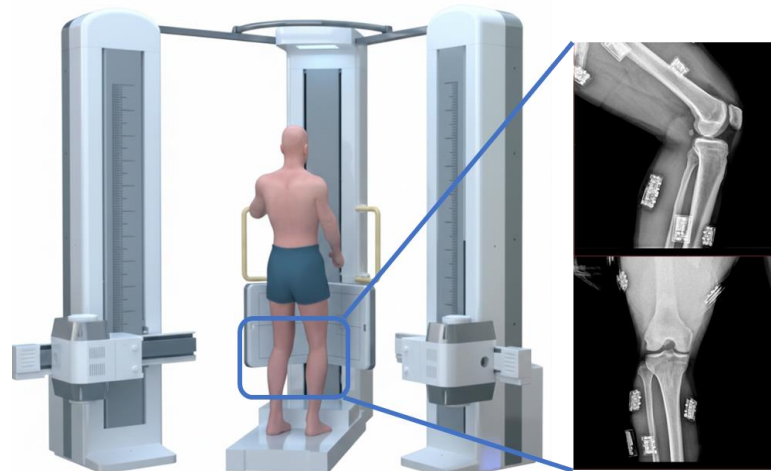


Figure 3. Dual-plane dynamic fluoroscopy system.

Compared with traditional methods, DFIS offers the combined advantages of in vivo, dynamic, and high-precision assessment. Studies have demonstrated its superior performance in eliminating STA: Gray et al. reported Root Mean Square Errors (RMSE) of only 0.78 mm for translation and 0.77° for rotation in knee joint analysis [23]; meanwhile, for the structurally more complex ankle joint, Cross et al. reported RMSE values as low as 0.59 mm and 0.71° [24, 25]. Consequently, this technology has established itself as a powerful tool for quantifying deep skeletal motion and joint contact kinematics.

1.1.2 Current Status of FEA Technology

The human lower limb comprises complex skeletal geometries and anisotropic soft tissue microstructures. Traditional modeling methods, such as Multibody Dynamics, typically treat bones as rigid bodies, thereby neglecting contact surface deformation and internal stress distribution. With the advancement of computational science, the FEM has become an indispensable simulation tool in the field of biomechanics. The core concept of FEM is discretization, which involves partitioning a continuous complex geometry into a finite number of regularly shaped small elements. By solving the equilibrium equations at the element nodes, the method simulates the mechanical response of the overall structure, including stress, strain, and displacement [26]. This method is capable not only of simulating complex geometric boundaries but also of handling nonlinear material properties and contact problems.

In the field of lower limb biomechanics, the application of FE technology primarily focuses on three areas. The first is injury mechanism and protection research, where detailed models of lower limb bones, joints, and soft tissues are established to simulate injury thresholds under extreme conditions such as falls, impacts, or car accidents. For instance, Untaru et al. developed a driver lower limb model to analyze the injury transmission mechanism in collision accidents, providing a basis for protective equipment design [27]. The second area involves clinical diagnosis and fracture risk assessment. Individualized FE models based on CT or MRI data [28] can predict fracture risk more

accurately than Bone Mineral Density alone, and the introduction of machine learning algorithms can further enhance the ability to identify high-risk populations [29]. The third area is implanting design and rehabilitation assessment. In joint replacement or internal fracture fixation, FEA is utilized to optimize the material, shape, and fixation position of implants, predicting the risk of stress shielding or loosening at the postoperative bone-implant interface [29]. Furthermore, in the design of rehabilitation aids such as exoskeletons and orthotic insoles, FEA contributes to optimizing structural lightweighting and interface pressure distribution [30].

However, the accuracy of FEA is highly dependent on the setting of boundary conditions and loading. Existing FE studies mostly adopt idealized motion boundaries or generic gait data, making it difficult to reflect the complex joint kinematics of specific individuals during real motion. Therefore, the inevitable trend for future development is the deep integration of high-precision in vivo motion capture data, such as that from DFIS, with high-fidelity FE models. Combined with rigorous validation through cadaveric experiments or clinical imaging data, this approach aims to build high-precision, subject-specific biomechanical simulation platforms [31].

1.2 Human Foot Biomechanics Research

The human foot is regarded as a masterpiece of engineering and natural evolution [32, 33]. Its unique skeletal arrangement forms a complex arched structure comprising the MLA, the Lateral Longitudinal Arch (LLA), and the Transverse Arch, as illustrated in **Figure 4**. This structure endows the foot with dual functional characteristics: during the landing and cushioning phase, it acts like a compliant spring for energy absorption and shock attenuation, adapting to uneven surfaces such as soft beaches; conversely, during the push-off and propulsion phase, it transforms into a rigid lever to effectively transmit Ground Reaction Forces (GRF) [34]. This stiffness regulation mechanism, characterized by the ability to switch between compliance and rigidity, lies at the core of foot function. Specifically, the dynamic stiffness variation of the MLA is considered crucial for regulating the mechanical performance of the entire foot [35, 36].



Figure 4. The arch structure of the human foot.

1.2.1 Foot Kinematics Research Based on Optical Motion Capture

Traditional research frequently employs optical motion capture systems combined with multi-segment foot models to analyze foot kinematics and stiffness. Raychoudhury et al. integrated infrared markers with three-dimensional force plates, simplifying the foot into two segments, the rearfoot and the forefoot, while modeling the tarsometatarsal joint as a hinge joint. Their study revealed significant discrepancies in position and orientation between the functional axis calculated via kinematics and the anatomical axis derived from anatomical landmarks. Specifically, the functional axis was consistently located superior and anterior to the anatomical axis [37], as depicted in **Figure 5a**. Boonpratong et al. further refined this approach by constructing a multi-joint model comprising the shank, rearfoot, forefoot, and toes, as illustrated in **Figure 5b** [38]. The research elucidated the mechanical transition mechanisms within the gait cycle. During the early to mid-stance phases, the foot primarily adapts to the ground through rotations of the talocrural and subtalar joints. Conversely, during the late stance phase, the involvement of the tarsometatarsal joint transforms this "two-hinge mechanism" into a "three-hinge mechanism," thereby significantly enhancing foot stiffness. Regarding foot arch deformation, Kelly et al. monitored the surface electromyography signals of intrinsic foot muscles, particularly the flexor hallucis brevis, flexor digitorum brevis, and abductor hallucis. They observed that as the load increased, the arch height decreased while its

length increased. However, upon activation of the intrinsic muscles, the arch length decreased and its stiffness significantly improved, demonstrating the critical role of active muscle contraction in maintaining foot arch stiffness [10].

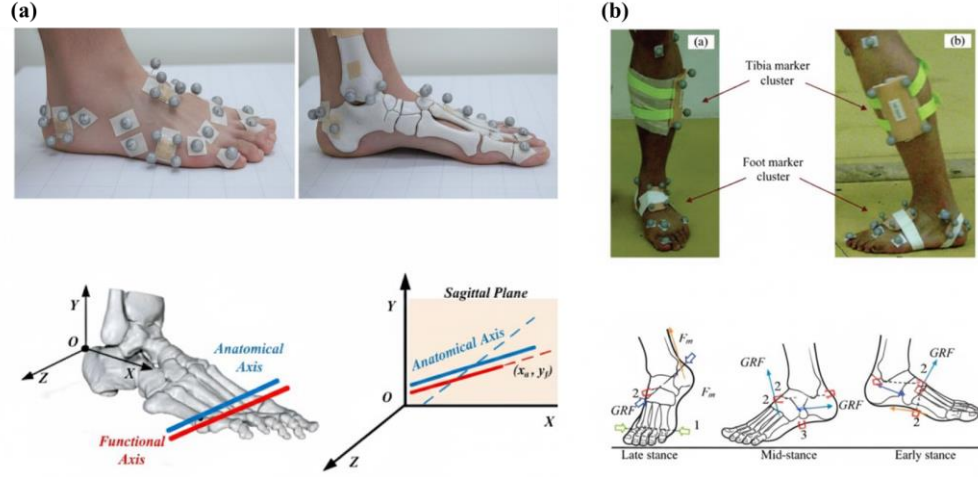


Figure 5. Multi-segment foot models: (a) The two-segment model by Raychoudhury et al.; (b) The multi-joint model by Boonpratong et al.

However, optical motion capture technology faces significant challenges in foot research. One major limitation is model simplification error, where the complex foot skeleton is reduced to merely three to five rigid segments [39], consequently ignoring the independent motion of numerous small joints. Another critical issue is the Soft Tissue Artifact. Nester et al. compared three tracking methods, skin-mounted markers, rigid plates, bone pins, and discovered that the first two non-invasive approaches exhibited substantial measurement errors in regions such as the talonavicular joint [40]. Although Maiwald et al. utilized invasive bone pin technology to acquire high-precision motion data of the tibia and calcaneus [41], the traumatic nature of this method precludes its widespread application in large-scale cohort studies or research on pathological gait.

1.2.2 Foot Kinematics Research Based on Dual-plane Fluoroscopy Technology

To address the precision limitations of optical motion capture, the DFIS has been introduced into foot research. However, the small size and dense arrangement of foot bones cause severe overlapping in X-ray projections, making image segmentation and registration extremely difficult [42]. Consequently, only a limited number of studies have

successfully utilized this technology to analyze the motion of joints associated with the MLA, such as the subtalar, talonavicular, and first metatarsophalangeal joints. Current research largely centers on elucidating the pathological mechanisms of flatfoot. Balsdon et al. employed a dual-plane system to compare the effects of barefoot walking, standard shoes, and orthotic insoles on the MLA, discovering that custom orthotics significantly increased the arch angle and altered its dynamic motion patterns [43]. Similarly, a study by Phan et al. on flatfoot patients revealed abnormal external rotation of the talus and arch collapse during the mid-to-late stance phase, alongside an increased Range of Motion (ROM) in multiple joints, indicating a significant reduction in joint stability [44]. Regarding normal feet, Phan et al. further observed that arch joints exhibit a coupled posture of "double-ended pronation and supination" during the late stance phase. This locking mechanism facilitates torque transmission during push-off [42]. Kelly et al. utilized this technology to investigate the role of the plantar fascia, finding a positive correlation between fascia stiffness and arch height. During the push-off phase, the stretching of the plantar fascia occurs synchronously with toe dorsiflexion. This phenomenon, known as the Windlass mechanism and illustrated in **Figure 6**, effectively mitigates arch collapse and enhances propulsion stability [45, 46].

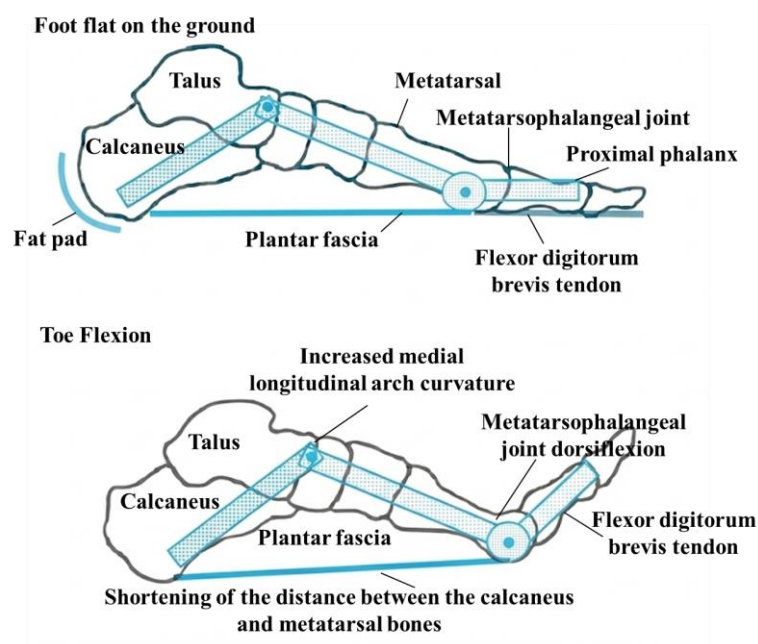


Figure 6. The Windlass mechanism of the foot.

In summary, although dual-plane technology has resolved the issue of soft tissue artifacts, existing studies are largely confined to the two-dimensional morphology of the MLA or the motion of single joints. There remains a lack of three-dimensional dynamic quantification of the Transverse Arch, and direct measurement of internal joint stress distribution is still not possible.

1.2.3 Current Status of Foot FEA Research

The FEM has become a pivotal tool in foot biomechanics due to its capability to simulate stress and strain distributions within biological structures efficiently and cost-effectively. Since Belytschko et al. first introduced FE techniques into orthopaedic biomechanics in 1973 [47], the technology has evolved from simplified models to high-fidelity representations. Tannous et al. integrated CAD with FE technology and introduced nonlinear constitutive relations for soft tissues. They successfully simulated stress transmission between ligaments and bones under various loads, thereby demonstrating the validity of FEA in predicting the mechanical response of biological tissues [48]. Similarly, Morales et al. simulated the contribution of lateral tendons to foot structural stability by applying virtual muscle forces. Although their model retained some simplifications in geometric details, it revealed the mechanical sensitivity associated with tendon synergy [49]. Furthermore, Zhang et al. [50] established a complete foot model for Hallux Valgus, as illustrated in **Figure 7**. A comparative analysis revealed a high stress concentration at the first metatarsophalangeal joint in patients, accompanied by reduced ligament contact force. These findings provide a mechanical explanation for the susceptibility of Hallux Valgus patients to joint pain and functional degeneration.

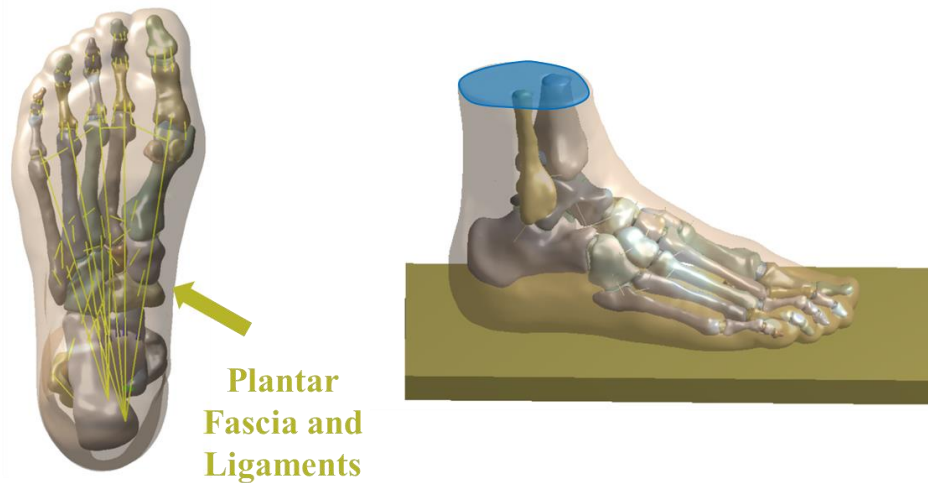


Figure 7. FE model of Hallux Valgus.

However, existing FE studies of the foot predominantly rely on static CT reconstructions and frequently employ simplified boundary conditions, such as assuming the ground to be a rigid flat plate or applying idealized loads. These models often lack integration with high-precision kinematic data derived from real gait patterns, a limitation that significantly constrains their predictive accuracy under dynamic operating conditions.

1.2.4 Combined Application of FEA and Dual-plane Imaging in Foot Research

The human foot possesses not only a flexible structure to adapt to the environment but also specific stiffness to propel the body forward. The unique arch structure of the foot plays a decisive role in this stiffness regulation [51]. Within this system, the Plantar Fascia (PF) serves multiple roles: it acts as a reservoir for elastic potential energy [52, 53] and functions as a critical structure for maintaining arch integrity [54]. The stiffness of the PF directly influences the dynamic regulation of the Windlass mechanism, thereby altering the morphology and function of the foot arch [55, 56]. Hicks [57] first proposed the concept of the Windlass mechanism: during the late stance phase of gait, as the heel rises, the metatarsophalangeal joint (MTPJ) undergoes passive dorsiflexion. This action pulls the PF tight, leading to calcaneal inversion and arch elevation. This synergistic action significantly enhances the propulsion efficiency of the foot. Cheng et al. [58] simulated this mechanism using the FEM and found that a greater MTPJ dorsiflexion angle resulted

in higher stress at the proximal attachment of the PF to the calcaneus. However, their study did not deeply investigate how variations in the stiffness of the PF itself reciprocally affect the Windlass mechanism and the internal stress distribution within the fascia. Furthermore, previous studies have frequently overlooked the influence of the Windlass mechanism on the curvature of the Posterior Transverse Arch. Research by Venkadesan et al. [59] demonstrated that through the coupling action of inter-metatarsal tissues, the transverse arch contributes more than 40% to the foot's longitudinal stiffness. The transverse arch not only protects plantar neurovascular structures from compression but also constitutes a reasonable elastic cushioning system for the foot [32, 60]. Therefore, when exploring the relationship between the Windlass mechanism and foot stiffness, the contributions of both the MLA and the rearfoot transverse arch must be considered simultaneously.

Based on this background, this study aims to establish an innovative framework combining the DFIS and FEA. The specific research objectives are as follows: To investigate the influence of PF stiffness. Building on the foundation laid by Cheng et al.[58], this study selects the critical moment of 30° MTPJ dorsiflexion during the stance phase of walking. By adjusting the Young's modulus of the PF in the FE model, different stiffness conditions are simulated. We will conduct an in-depth analysis of how changes in fascia stiffness impact the efficacy of the Windlass mechanism, the morphology of the MLA, and the dynamics of the rearfoot transverse arch.

In FE simulation, model accuracy is paramount. Traditional studies predominantly rely on plantar pressure or navicular drop for validation [61, 62]. In contrast, this study utilizes the high-precision DFIS to acquire in vivo motion data, directly measuring changes in longitudinal arch length and transverse arch width during foot motion [63]. This validation method, based on dynamic skeletal geometric parameters, effectively compensates for the inability of traditional optical capture to quantify internal skeletal micro-motion, providing a more targeted validation standard for the biological fidelity of the FE model. Through this research, we hypothesize that as PF stiffness increases, the curvatures of both the longitudinal and transverse arches of the foot will increase during

the engagement of the Windlass mechanism. This study expects to further reveal the energetic mechanisms of the human foot and provide theoretical references for the design of high-performance assistive devices, such as orthotic insoles and prosthetics.

1.3 Current Status of Human Knee Biomechanics Research

1.3.1 Evolution of Kinematic Measurement Methods for the Knee Joint

The knee joint represents one of the most complex diarthrodial joints in the human body. Its anatomical morphology determines that it is not a simple hinge joint but rather involves coupled motion with six degrees of freedom, encompassing sliding, rolling, and rotation. For a long time, researchers have proposed various theoretical models to describe its kinematic characteristics, evolving from the two-dimensional Instantaneous Center of Rotation [64] to the three-dimensional Instantaneous Helical Axis [10]. Correspondingly, measurement techniques have undergone a significant evolution from in vitro to in vivo, from invasive to non-invasive, and from low to high precision.

Early Radiostereometric Analysis experienced rapid development during the 1980s. This technique involves surgically implanting tantalum beads with a diameter of approximately 0.8 mm into the bone and utilizing X-rays to track the movement trajectory of these beads, thereby precisely reconstructing the three-dimensional motion of the skeleton [18]. Although Radiostereometric Analysis offers extremely high precision, its invasive nature limits its application in healthy populations and routine clinical testing. Subsequently, optical motion capture technology based on Skin Markers became the mainstream approach, as illustrated in **Figure 8**. This method is user-friendly and capable of assessing various activities such as walking and running. However, due to the presence of the Soft Tissue Artifact, or STA, the movement of markers on the skin surface fails to accurately reflect the true motion of the underlying deep bone, leading to compromised measurement accuracy [12]. Furthermore, while traditional CT or MRI scans can reconstruct high-precision three-dimensional anatomical models based on grayscale values, they are generally limited to static scanning in a supine position. This limitation

neglects the influence of gravity, ground reaction forces, and muscle contraction on joint alignment, thus failing to reflect true kinematics under physiological loads.



Figure 8. Optical motion capture technology based on skin markers.

In recent years, the Dual-plane Fluoroscopy System has been widely applied in knee joint research as a breakthrough kinematic assessment technology [65]. This technology combines non-invasiveness with high precision at the sub-millimeter level, effectively addressing the limitations associated with the methods. Researchers have utilized this technology to precisely quantify axial variations in the knee joint during dynamic flexion and extension [66], as illustrated in **Figure 9**. Descriptions of the knee motion axis frequently involve the Geometric Center Axis, known as GCA, and the Transepicondylar Axis, referred to as TEA. The GCA is defined as the line connecting the centers of spheres fitted to the medial and lateral femoral condyles, whereas the TEA represents the line connecting the anatomical landmarks of the medial and lateral epicondyles. Research confirms that the GCA exhibits superior stability compared to the TEA when serving as the rotational axis for knee flexion and extension movements [67]. Furthermore, dual-plane fluoroscopy technology has been employed in soft tissue morphological studies. For instance, it allows for the indirect calculation of ligament strain by measuring changes

in the distance between ligament attachment sites, thereby elucidating the biomechanical impact of flexion and extension movements on ligaments [67].

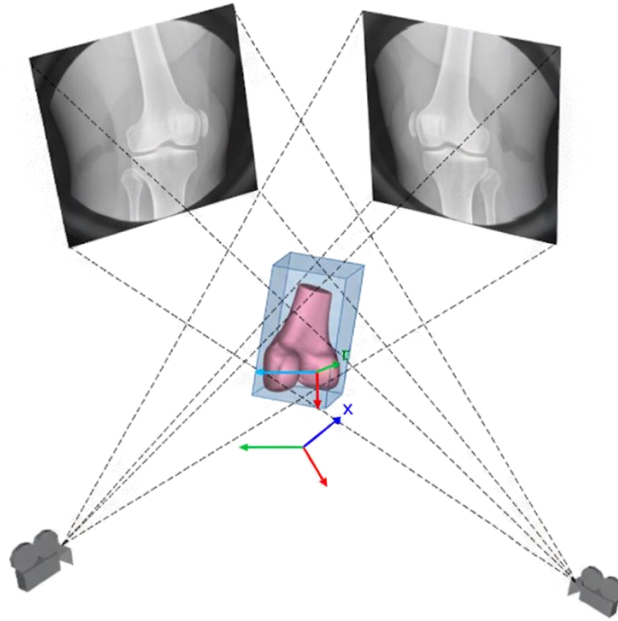


Figure 9. Knee joint motion trajectories acquired using the Dual-plane Fluoroscopy System.

1.3.2 FE Modeling Strategies for the Knee Joint

The biomechanical response of the knee joint is intrinsically linked to its kinematic state; therefore, the precise simulation of joint motion represents a core challenge in FEA. The establishment and analysis of the knee model are illustrated in **Figure 10**. Early FE studies were predominantly based on static models at single flexion angles, rendering them incapable of reflecting mechanical changes during continuous motion [68]. Although some studies considered multiple angles, they simulated flexion and extension by simply rotating the femur or tibia. This approach ignored the complex coupled rolling-sliding mechanism of the knee, leading to compromised simulation accuracy [69]. To simulate authentic motion, researchers have primarily adopted two driving strategies.

The first strategy involves the Force-driven, or Kinetics-based, model, which emphasizes dynamics. Some studies utilized cadaveric experiments to obtain contact force data for loading the model [70]. A more advanced approach combines optical motion capture with

Musculoskeletal Modeling. By employing inverse dynamics, researchers calculate muscle forces and joint reaction forces to serve as boundary conditions driving the FE model [71]. The primary advantage of this method lies in its consideration of active muscle contraction.

The second strategy is the Displacement-driven, or Kinematics-based, model, which directly utilizes kinematic data to control skeletal positioning. Compared to force-driven models, this modeling process is relatively straightforward and circumvents the uncertainties associated with complex muscle force estimation. However, data sources for traditional displacement-driven models often originate from the step-wise registration of static CT or MRI scans [72], or they simplify translational motion by considering only primary degrees of freedom, such as flexion and extension [73].

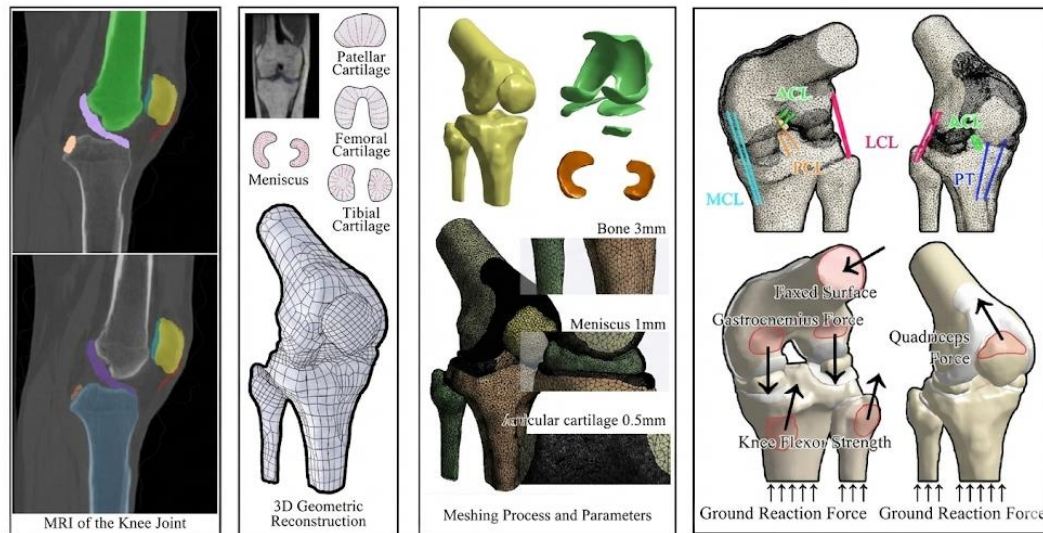


Figure 10. Establishment and analysis of the knee FE model.

Consequently, the absence of high-precision continuous in vivo motion data directly constrains the accuracy of FE models. At present, there is a paucity of research that directly utilizes 6-DOF real-time motion data obtained from the DFIS as displacement boundary conditions, while simultaneously integrating muscle forces calculated by musculoskeletal models to drive FEA. Addressing this technical bottleneck is precisely the breakthrough this study aims to achieve.

1.3.3 Combined Application of FEA and Dual-plane Imaging in Knee Joint Research

Stair ascent represents a daily activity that places exceptionally high demands on the knee joint. Compared to level walking, stair ascent involves greater knee flexion angles and more intense muscle contractions, thereby subjecting the menisci and articular cartilage to elevated shear and compressive forces. This significantly increases the risk of degenerative joint pathologies, particularly among the female and elderly populations. However, stair ascent involves multi-degree-of-freedom compound motions, time-varying muscle forces, and fluctuating ground reaction forces. Traditional three-dimensional kinematic analyses or static FE models are unable to capture these complex real-time load transmission mechanisms. Furthermore, static or idealized models frequently overlook authentic motion trajectories and individualized muscle coordination patterns, thereby limiting their clinical relevance.

The knee joint is one of the most heavily loaded and structurally complex joints in the human body, playing a crucial role in daily and competitive activities such as walking and jumping. Within this structure, the PT serves as a critical mechanical link connecting the quadriceps to the tibia, responsible for transmitting immense muscle forces during knee extension and dynamic buffering [74]. However, this high-intensity mechanical load makes the PT highly susceptible to overuse injuries. Statistics indicate that the prevalence of "Jumper's Knee," or patellar tendinopathy, reaches as high as 45% in volleyball players and 32% in basketball players [75]. Repetitive overloading is generally considered the primary etiological factor leading to PT pathology [76-78]. Consequently, a profound understanding of the biomechanical characteristics of the PT during high-impact activities such as jump landings, particularly the magnitude and spatial distribution of internal stress is vital for formulating precise prevention and rehabilitation strategies.

Considering the challenges, this study established an innovative biomechanical framework integrating Musculoskeletal Modeling, the DFIS, and FEA. This framework aims to investigate the internal mechanical mechanisms of the knee joint under dynamic conditions such as landing and stair ascent. Specifically, DFIS provided three-

dimensional skeletal motion trajectories with high spatiotemporal resolution, ensuring the kinematic authenticity of the FE model boundary conditions. Concurrently, the musculoskeletal model calculated individualized dynamic muscle forces, providing precise load boundary conditions for the analysis. Finally, based on these accurate motion and load inputs, FEA calculated the stress fields within the PT, menisci, and cartilage.

Compared with quasi-static methods that apply standardized boundary conditions, the approach utilized in this study fully accounts for individual differences, variations in muscle forces, and real joint angles during motion. Consequently, it reproduces the load-bearing state of the knee joint with greater physiological realism. Furthermore, this study integrated statistical learning methods, such as Ridge Regression, to explore simplified indicators for predicting peak equivalent stress in the PT. These findings are expected to provide theoretical grounds and methodological support for real-time monitoring, early warning systems, and rehabilitation training for knee injuries.

2. MATERIAL AND METHODS

2.1 Overview of the experimental and computational workflow

This study aims to establish a lower limb biomechanics research framework that integrates the DFIS, gait analysis, and FEA. The overall research workflow is divided into three core stages: Data Acquisition, Modeling & Reproduction, and Simulation & Analysis. By combining high-precision in vivo skeletal kinematics with kinetics calculated by musculoskeletal models, the framework provides individualized geometric boundaries and loading conditions for the FE model. Based on this framework, biomechanical investigations targeting the foot and knee were conducted, as illustrated in **Figure 11**.

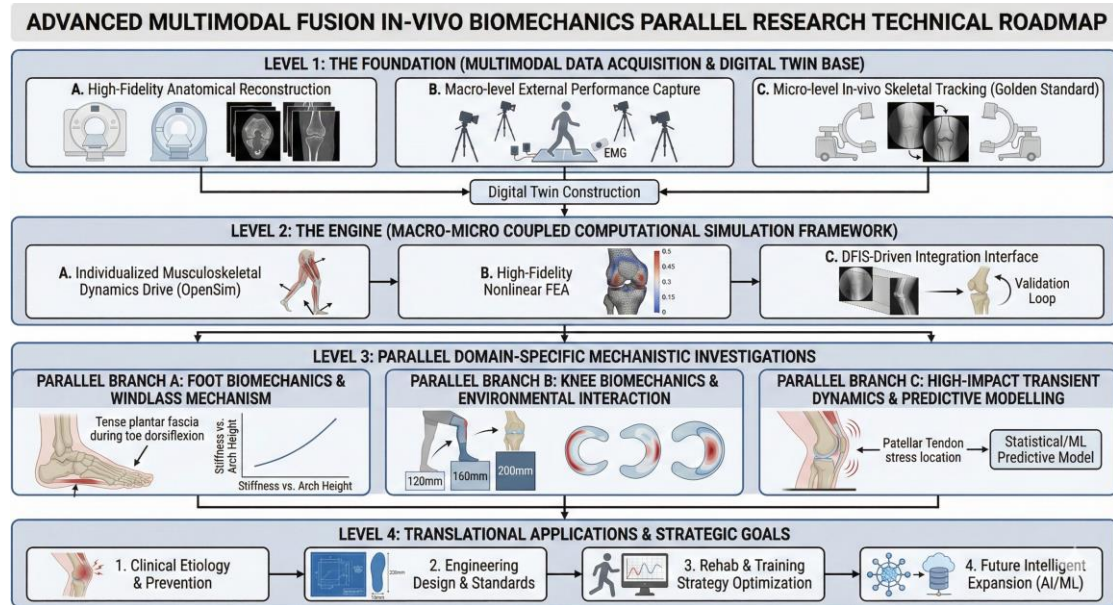


Figure 11. Biomechanical study of the foot and knee joints based on FEA and the DFIS

For the foot study, the research focuses on investigating the influence of PF stiffness on the Windlass Mechanism and the structural function of the foot arch. Subjects performed natural walking at a speed of 5 km/h. The DFIS was utilized to accurately capture skeletal motion during locomotion, allowing for the direct measurement of dynamic changes in the length and height of the MLA and the width and height of the Rearfoot Transverse Arch. These in vivo measurements served as the "gold standard" to validate the geometric prediction accuracy of the FE model. Building upon the validated model, the material

properties of the PF were systematically altered to simulate different stiffness conditions. This allowed for a quantitative analysis of how varying levels of stiffness contribute to the maintenance of foot arch morphology and the distribution of internal stress.

For the knee joint study, a Quasi-dynamic FE Model driven by real motion was constructed, encompassing two primary application scenarios, Stair Ascent: An analysis of the impact of varying stair heights on the stress distribution within the knee menisci and articular cartilage during stair climbing. Jump Landing: Individualized multi-moment modeling targeting the PT was conducted to characterize internal tendon stress under high-impact loads.

To achieve this, the 6-DOF knee motion acquired via DFIS was employed as the displacement boundary condition, while relevant muscle forces estimated by the Musculoskeletal Model served as force boundary conditions. This approach achieved a deep coupling of kinematics and kinetics.

2.2 Instruments and Equipment

To achieve the objectives, this study integrated various state-of-the-art experimental devices and simulation software. The specific workflow is detailed as follows:

Static Anatomical Imaging: A Somatom Sensation 16-slice spiral CT scanner (Siemens, Germany) was utilized to acquire high-resolution static images of the subjects' lower limbs. These images served as the foundation for constructing three-dimensional geometric models ranging from bone to soft tissue.

Dynamic Fluoroscopic Imaging: A DFIS (Taoimage, Shanghai, China) was employed to collect dynamic X-ray images during motion at a sampling rate of 30–130 Hz (depending on specific experimental settings). Images extracted at critical moments of the stance phase were used for 2D-3D skeletal motion registration.

Gait Data Acquisition: A Vicon three-dimensional motion capture system (Oxford Metrics, UK) was used to synchronously collect the trajectories of surface markers.

Ground Reaction Force (GRF) Acquisition: A Kistler 3D force plate (Kistler, Switzerland) was utilized to record ground reaction forces.

Electromyography (EMG) Data Acquisition: A Delsys wireless surface EMG system (Delsys, Boston, MA, USA) was used to record muscle activation signals from the rectus femoris, biceps femoris, and gastrocnemius. These data assisted in validating the muscle force predictions generated by the musculoskeletal model.

Modeling and Simulation Software: Geometry Reconstruction: CT data were imported into Mimics 21.0 (Materialise, Belgium) for threshold segmentation and three-dimensional reconstruction to generate .stl files. Subsequently, reverse engineering processing (e.g., surface smoothing, denoising) was performed in Geomagic Studio (Geomagic, USA). Finally, solid models of the meniscus, cartilage, and ligaments were constructed in SolidWorks 2020 (SolidWorks Corp., USA) based on anatomical references.

Musculoskeletal Modeling: Vicon and Kistler data were imported into the generic musculoskeletal model (Gait 2392) within OpenSim (Stanford University, USA). Muscle forces were estimated through scaling and inverse dynamics solutions.

FEA: The geometric models, kinematic boundaries (derived from DFIS), and mechanical boundaries (derived from OpenSim) were integrated into ANSYS Workbench 2021 R1 (Ansys, Inc., USA) to perform meshing, material assignment, and nonlinear FE solutions.

Detailed information regarding experimental testing and FE simulation is provided in Sections 2.5 and 2.6. From a technical perspective, this study establishes an individualized lower limb FE modeling workflow based on dual-plane fluoroscopy. This provides a high-precision, minimally invasive biomechanical research method for accurately acquiring skeletal kinematic information of the foot and knee. Furthermore, by utilizing the validated foot FE model to quantitatively analyze the influence of PF stiffness on the Windlass mechanism and arch stability, this research contributes to a deeper mechanical understanding of the pathogenesis of foot-related diseases and intervention strategies. Finally, leveraging the quasi-dynamic FE model of the knee, this

study reveals the stress response characteristics of the meniscus and PT under varying stair heights and jump landing conditions, providing a reliable biomechanical basis for injury prevention, rehabilitation optimization, and safety assessment in daily activities.

2.3 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**.

A

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: _____

Figure 12. Human informed consent form

2.4 Participants

This study comprises three independent experimental components: foot biomechanics experiments, and knee biomechanics experiments involving stair ascent and jump landing. All subjects were free from any history of lower limb musculoskeletal disorders and provided written informed consent prior to participation. This study was approved by the Research Ethics Committee of Ningbo University (No. RAGH20230124).

2.4.1 Foot Biomechanics Experiment

A healthy male volunteer (age: 26 years; height: 181 cm; weight: 80 kg) was recruited for this study. The subject had no history of musculoskeletal pathology or lower limb injury. To ensure the representativeness of the sample, the Arch Height Index (AHI) [79] and Foot Posture Index (FPI) [80] were employed to classify the subject's foot type [81]. The assessment results indicated that the subject's right foot exhibited an AHI > 0.356 and an FPI between -4 and 4, meeting the criteria for a Normal Arch and Neutral Posture.

Regarding data acquisition, as illustrated in **Figure 13**, the subject performed natural walking at a speed of 5 km/h. The DFIS was utilized to capture in vivo skeletal motion of the foot, with a specific focus on measuring the length and height of the MLA, as well as the width and height of the Rearfoot Transverse Arch. Synchronously, a Vicon three-dimensional motion capture system (Oxford Metrics Ltd., UK) was employed to acquire kinematic data. This data was subsequently imported into OpenSim (Stanford University, USA) to construct a musculoskeletal model, thereby providing boundary conditions for the subsequent FEA.

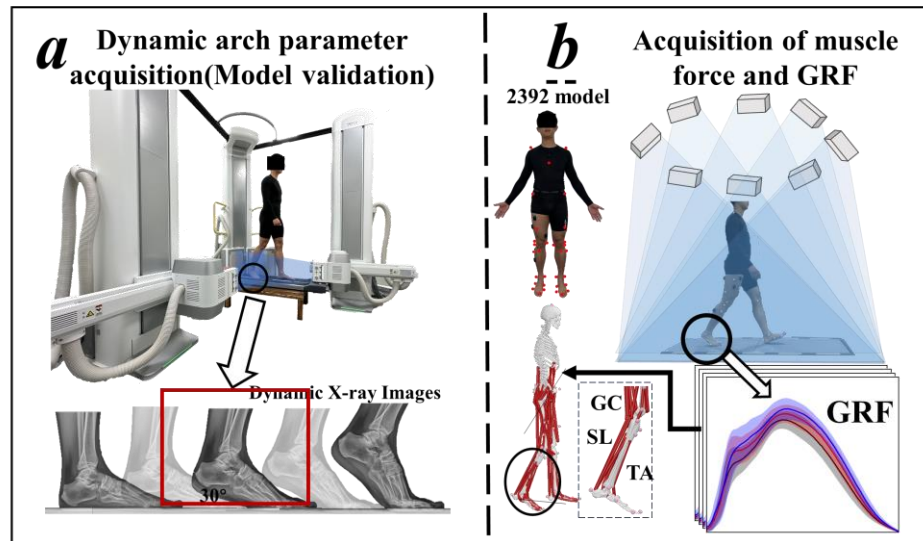


Figure 13. Acquisition of foot imaging, kinematic, and kinetic data from the subject

2.4.2 Knee Biomechanics Experiment - Stair Ascent Condition

A male subject (age: 25 years; height: 172.8 cm; weight: 70.5 kg) was recruited for this experiment. His anthropometric data aligned closely with the 50th percentile of Chinese adult males [82], ensuring that the subsequently constructed FE model possessed typical geometric representativeness. Exclusion criteria included: no history of lower limb surgery within the past six months, and no history of lumbar, pelvic, or systemic diseases, neuromuscular dysfunction, or balance impairments [83].

The objective of this experiment was to investigate the influence of different step heights on the stress distribution of the knee joint during the single-leg stance phase of stair ascent, as illustrated in **Figure 14**. The experiment utilized a custom-built seven-step staircase apparatus with a fixed step depth of 180 mm. By adjusting the step height, five distinct experimental conditions were established (120 mm, 140 mm, 160 mm, 180 mm, and 200 mm) to systematically study the effects of varying step heights on the biomechanical characteristics of the knee joint.

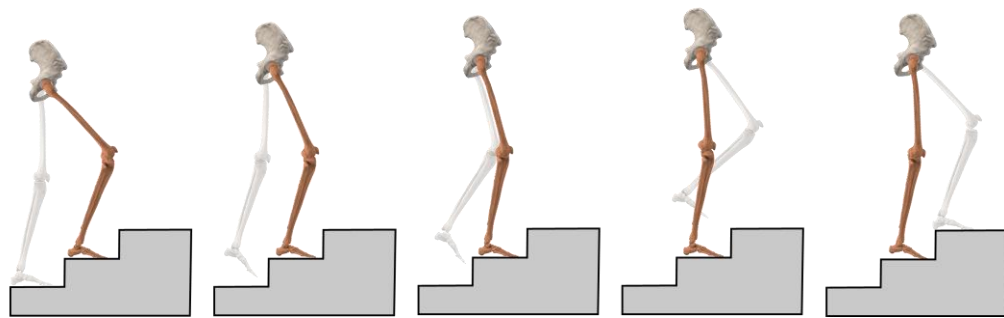


Figure 14. Single-leg stance phase during stair ascent

2.4.3 Knee Biomechanics Experiment - Jump Landing Condition

A young male subject (age: 27 years; height: 178 cm; weight: 68 kg) was recruited for this experiment. The subject was free from any history of neuromuscular disorders or biomechanical abnormalities. Based on the Crossed Symmetry principle[84], the subject's dominant leg (defined as the lower limb contralateral to the shooting arm) was selected as the primary focus of analysis to investigate the mechanical response of the knee joint during the jump landing process.

2.5 Acquisition of Muscle Force

This study employs a musculoskeletal modeling-driven approach to analyze muscle kinetic parameters across various movement conditions. By synchronously acquiring kinematics, kinetics, and electromyography (EMG) signals, muscle forces are calculated and validated to provide precise loading boundary conditions for the FE Model.

2.5.1 Acquisition of Foot Muscle Forces during Walking

A Vicon motion capture system (Oxford Metrics, UK) equipped with 10 infrared cameras and an embedded Kistler 3D force plate (Kistler, Switzerland) were employed to synchronously acquire lower limb marker trajectories and GRF at sampling frequencies of 200 Hz and 1000 Hz, respectively. The subject performed walking trials at a speed of 1.4 ± 0.2 m/s (approximately 5 km/h). Ten valid sets of kinematic and kinetic data were recorded and averaged for subsequent analysis [85]. The acquired data were imported into the generic musculoskeletal multi-body model (Gait 2392 Model) within OpenSim software to perform Inverse Dynamics calculations for estimating muscle forces[86]. This section specifically extracted three core muscle groups governing foot motion: the Gastrocnemius (GC) and Soleus (SL), which serve as the primary power sources for ankle plantarflexion, and the Tibialis Anterior, which is the principal muscle responsible for generating ankle dorsiflexion moments during the gait cycle [87].

2.5.2 Acquisition of Muscle Forces during Stair Ascent

In the stair ascent experiment, the Vicon three-dimensional motion capture system (200 Hz) was similarly utilized to acquire kinematic data. To obtain kinetic data during the stance phase of the right leg, a portable Kistler 3D force plate (1000 Hz) was securely mounted on the surface of the fourth step of the staircase apparatus[88-90]. The subject's average ascent speed was controlled at 0.88 ± 0.19 m/s. Utilizing the Gait 2392 model within OpenSim [91], lower limb muscle forces were calculated based on the kinematic and kinetic data obtained under various step heights, as shown in **Figure 15**. This section specifically focused on the analysis of the Quadriceps, Hamstrings, and Gastrocnemius

[92]. Ultimately, the calculated muscle forces were integrated with the GRF to serve as external loads and boundary conditions applied to the knee FE model.

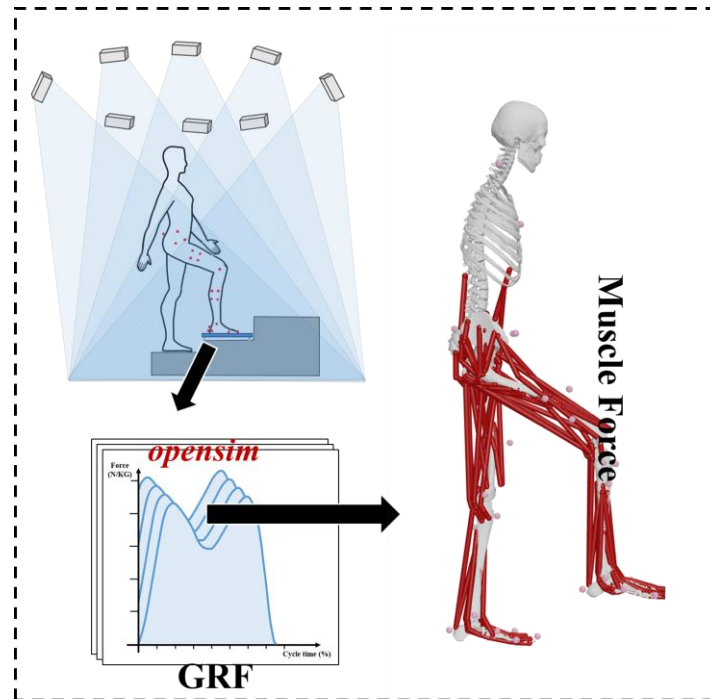


Figure 15. Calculation of lower limb muscle forces based on the OpenSim model

2.5.3 Acquisition of Knee Muscle Forces during Jump Landing

The subject first performed a standardized 5-minute dynamic warm-up. Subsequently, two maximal-effort stop-jumps were executed to touch a height marker, and 85% of the average height achieved was established as the target height for the formal experiment. A successful trial required the subject to touch the target marker with their right hand and land with both feet on the force plates [93]. A 10-camera Vicon system (200 Hz) captured motion trajectories, while two embedded Kistler force plates (1000 Hz) synchronously recorded GRF, as illustrated in **Figure 16**. Data from the moment of Initial Contact to Maximum Knee Flexion were selected for analysis. A Delsys wireless surface EMG system (Delsys, Boston, MA, USA) synchronously recorded muscle signals from the right lower limb at a frequency of 1000 Hz, including the Rectus Femoris, vastus lateralis, Vastus Medialis, Biceps Femoris, and Gastrocnemius [94].

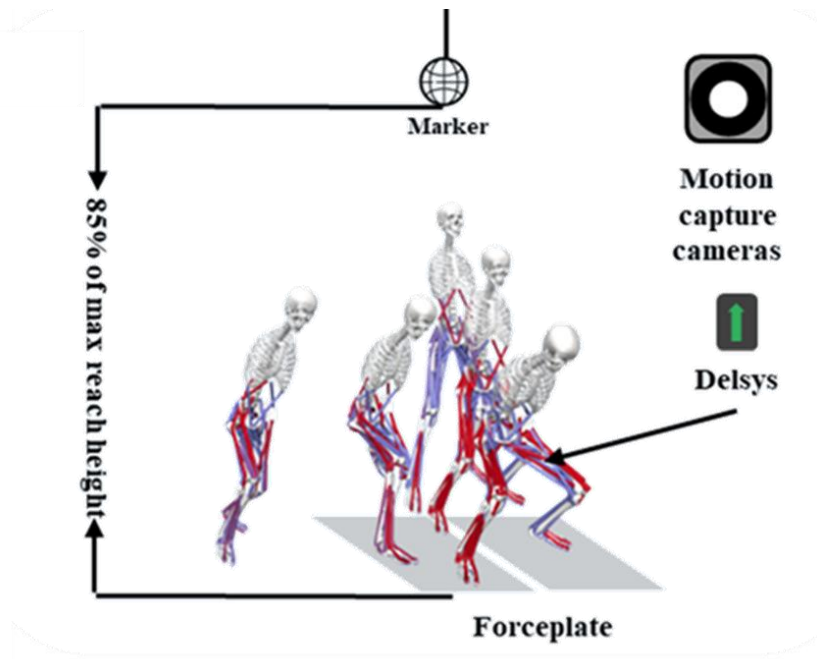


Figure 16. Acquisition of kinematic and kinetic data during landing to obtain muscle forces

2.2.3 Processing of Electromyography Signals

The raw EMG signals were imported into Delsys EMGworks Analysis 4.2 software for processing. First, a fourth-order Butterworth band-pass filter (10–500 Hz) was applied to eliminate noise and artifacts [95, 96]. Following baseline correction and full-wave rectification, the Root Mean Square values were calculated. The EMG data were then normalized using the subject's Maximum Voluntary Contraction to obtain muscle Activation Levels ranging from 0 (no activation) to 1 (100% activation) [97-100]. Finally, the experimentally measured muscle activation levels were compared with those predicted by the OpenSim Static Optimization algorithm to validate the accuracy and biological fidelity of the musculoskeletal model [101].

2.6 Establishment of FE Models

2.6.1 Establishment of the Foot FE Model

The foot model is established based on the subject's foot imaging CT data. The foot model consists of 29 bones, one PF, five Metatarsophalangeal articular capsules, four deep

transverse metatarsal ligaments, 66 ligaments and one wrapped soft tissue. The contact between bone and cartilage surface is set as frictionless, and there is friction between foot and plate, and the friction coefficient is 0.2, as shown in **Figure 17**. In all meshing, the plate is divided into 4mm, and the maximum von Mises stress of different tissues in the FE model is calculated and compared. When the difference between the predicted results of the two mesh schemes is less than 5% [102], it is considered that the mesh is convergent, and the final mesh size and material properties is shown in **Table 1**. All materials used in this FE analysis are assumed to be homogeneous, isotropic and linear elastic. Elastic properties are defined by two material constants: Young's modulus (E) and Poisson's ratio (ν) [103]. In particular, the Young's modulus of 350Mpa is selected as the reference value to represent the regular PF stiffness. At the same time, in order to evaluate the effect of PF stiffness on stress distribution, eight different Young's modulus values ranging from 87.5 to 700Mpa (Equivalent to 10% to 200% of the reference value) were assigned to PF [104].

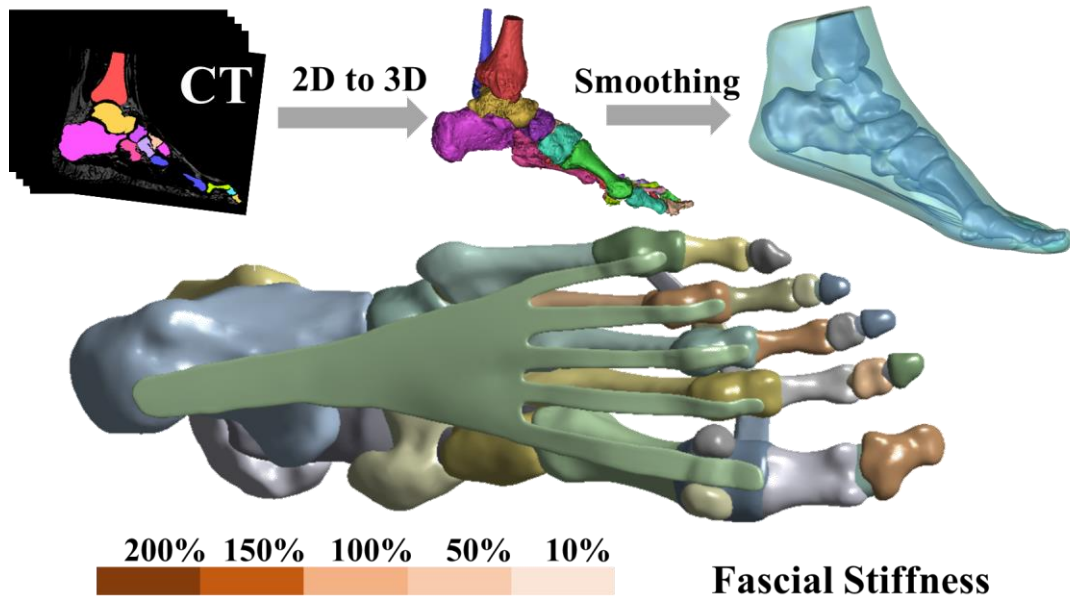


Figure 17. Establishment of the Foot FE Model

2.6.2 Establishment of the Knee FE Model

A CT scan was performed to acquire detailed imaging data of the participant's right knee joint, using a Somatom Sensation 16-slice spiral CT scanner (slice thickness: 0.5 mm; 120 kV; 250–300 mA)[105]. Tibial, femoral, fibular, and patellar structures were manually segmented from the CT images using the 3D reconstruction software Mimics 21.0 (Materialise, Leuven, Belgium). The segmented data were saved as 3D models (. stl format) and then imported into Geomagic (Geomagic Inc., Research Triangle Park, NC, USA) for surface smoothing [106]. Subsequently, cartilage, ligaments, and menisci were modeled in SolidWorks 2020 based on anatomical references [60], as shown in **Figure 18**, the final 3D FE model of the knee joint included four bones, five cartilage layers, five ligaments, and both medial and lateral menisci.

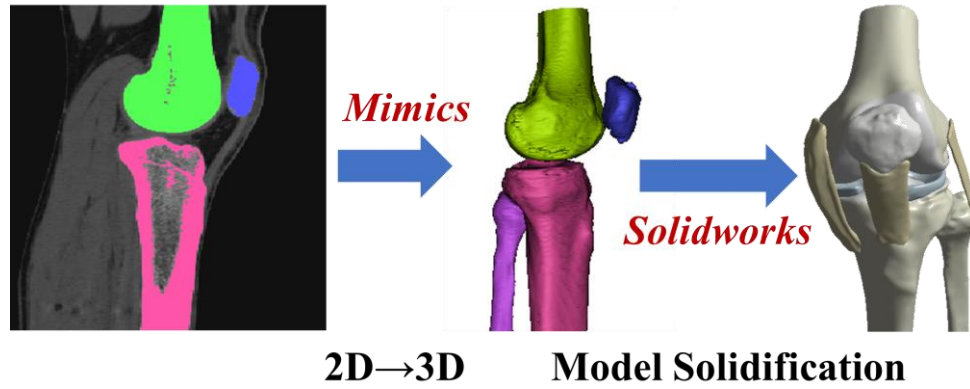


Figure 18. Establishment of the Knee FE Model

2.6.3 Material Properties of the Foot FE Model

All materials used in this FE analysis are assumed to be homogeneous, isotropic and linear elastic. Elastic properties are defined by two material constants: Young's modulus (E) and Poisson's ratio (ν) [103]. In particular, the Young's modulus of 350Mpa is selected as the reference value to represent the regular PF stiffness. At the same time, in order to evaluate the effect of PF stiffness on stress distribution, eight different Young's modulus values ranging from 87.5 to 700Mpa (Equivalent to 25% to 200% of the reference value) were assigned to PF [104], the material properties of the foot shown in **Table 1**.

Table 1. Material properties, mesh size of each component and values of muscle force

Component	Young's Modulus E (MPa)	Poisson's Ratio ν	Mesh Size(mm)	Force (N)
Bone	7300	0.3	2mm	/
Soft tissue	1.15	0.49	3mm	/
Cartilage	1	0.4	1mm	/
Ligament	260	0.4	/	/
Planter Fascia	350 (87.5-700)	0.4	1mm	/
Ground plate	17000	0.1	4mm	/
GRF	/	/	/	924.34
GC	/	/	/	52.24
SL	/	/	/	83.54
TA	/	/	/	14.24

Note: GRF: ground reaction force, GC: gastrocnemius muscle, SL: soleus muscle, TA: tibialis anterior muscle

2.6.4 Material Properties of the Knee FE Model

The FEA was conducted using ANSYS Workbench 2021 R1 (Ansys, Inc., Canonsburg, PA, USA) to evaluate the stress distribution. To ensure the accuracy and reliability of the model, all solid components were meshed using tetrahedral elements [107]. A mesh convergence study was performed, resulting in mesh sizes of 3 mm for bones, 1 mm for encapsulated soft tissues, and 0.5 mm for cartilage [108]. The material properties used are summarized in the **table 2** [109-111].

The ligaments were modelled as transversely isotropic nearly incompressible neo-Hookean materials[112-115] using the strain energy function, as in Eq. (1).

$$\Phi = C_1(\overline{G_1} - 3) + \frac{1}{D_1}(J_F - 1)^2 + S(\lambda) \quad (1)$$

Here, $S(\lambda)$ represents the strain energy function of the fibre family which satisfies the following conditions, as in Eq. (2).

$$\lambda \frac{dS}{d\lambda} = \begin{cases} 0, \lambda \leq 1 \\ C_3(e^{(\lambda-1)C_4}) - 1, 1 < \lambda < \lambda^* \\ C_5\lambda + C_6, \lambda \geq \lambda^* \end{cases} \quad (2)$$

C_1 is a bulk material constant related to the shear modulus μ ($C_1 = 2/\mu$). J_F is the Jacobian matrix of the deformation gradient F , and $\overline{G}_1$ denotes the first invariant of the left Cauchy–Green tensor $\overline{G}_1 = \text{tr} \overline{F} \overline{F}^T$ with the modified deformation gradient $\overline{F}$ ($\overline{F} = J_F^{-0.33} F$).

The stress in the fibres was dependent on the fibre stretch λ , which is determined from the deformed fibre orientation a_d , the deformation gradient F , and the initial fibre orientation a_0 ($\lambda \cdot a_d = F \cdot a_0$). Under compression of $\lambda \leq 1$, the fibres did not support any compressive stresses. It is noteworthy that the stiffness of the fibres increased exponentially when the fibre stretch was within the range from 1 to the predefined value (λ^*). Beyond this stretch, the fibres straightened and the stiffness increased linearly. The initial fiber orientation a_0 was specified according to the geometry of the local element. In the ligaments, the fibers were oriented in line to the principal axis of the geometry.

The constant C_3 scales the exponential stress, C_4 is related to the rate of collagen uncramping, and C_5 represents the elastic modulus of the straightened collagen fibres. The constant C_6 was introduced to ensure stress continuation at λ^* [$C_6 = (e^{(\lambda^*-1)C_4} - 1)C_3 - (C_5\lambda^*)$]. The values of the material constants C_1 , C_3 , C_4 , C_5 , and D_1 are listed in **table 2** [116].

Table 2. Knee joint FE model assumptions, material assignments and element types

Structure		Element Count	Material Properties	
			Young's Modulus (MPa)	Poisson's Ratio
Bone	Femur	83541	14200	0.3
	Tibia	60452		
	Fibula	102284		
	Patella	8517		
Cartilage	Femoral Cartilage	10256	15	0.46

	Tibial Cartilage	30038				
	Patellar Cartilage	50245				
	Tibiofibular Cartilage	9414				
Meniscus	Medial Meniscus	40453				
	Lateral Meniscus	22571		27.5	0.33	
	$C_1(\text{MPa})$	$C_3(\text{MPa})$	$C_4(-)$	$C_5(\text{MPa})$	$D_1(\text{MPa}^{-1})$	$\lambda^*(-)$
ACL	1.95	0.0139	116.22	535.039	0.00683	1.046
PCL	3.25	0.1196	87.178	431.063	0.0041	1.035
LCL	1.44	0.57	48.0	467.1	0.00126	1.065
MCL	1.44	0.57	48.0	467.1	0.00126	1.063
PT	3.25	0.1196	87.178	431.063	0.0041	1.035

Note: MCL: Medial Collateral Ligament, LCL: Lateral Collateral Ligament, PT: Patellar Tendon, ACL: Anterior Cruciate Ligament, PCL: Posterior Cruciate Ligament

Frictionless contact was defined between the menisci and both the tibial and femoral cartilages, while bonded contact was assigned between each cartilage and its corresponding bone [93]. The distal ends of the tibia and fibula were constrained in all six degrees of freedom. The average muscle forces of the quadriceps, biceps femoris, and gastrocnemius muscles across all datasets were computed and applied to the model. A vertically downward force was also applied to the proximal femur to simulate body weight.

2.6.5 Foot Loading and Boundary Conditions

In this study, the boundary conditions include both prescribed kinematics and force. We prescribed the posture related kinematics 30° dorsiflexion at the first MTP joint with the center of rotation at the metatarsal head and constrained the ground plate to have only a vertical DOF. These constraints define the relative configuration of the foot with respect to the ground during push-off, but they do not define the complete final state of the system. In particular, the final distributions of plantar fascia tension and arch deformation were

not specified a priori; they emerged from the FE computation considering nonlinear tissue mechanics, contact/friction, and muscle forces . The ground-plate vertical displacement was not assumed to be known beforehand; instead, it was iteratively updated in a displacement-controlled procedure until the simulated reaction force converged to the experimentally measured vGRF. Therefore, only selected kinematic parameters were prescribed, while the final mechanical response was solved rather than imposed.

To accurately simulate the Windlass Mechanism and assess its impact on PF tension and foot arch morphology, this study avoided the simplified boundary conditions often employed in traditional FE analyses, where the tibia is completely fixed relative to the ground. Instead, the angle between the foot and the ground was adjusted to maintain a dorsiflexion angle of 30° at the first MTPJ, with the center of rotation defined at the metatarsal head. This configuration was designed to simulate the specific posture during the push-off phase when the Windlass mechanism is activated. Furthermore, precise muscle forces derived from the musculoskeletal model, specifically the gastrocnemius, soleus, and tibialis anterior, were applied to their respective bony attachment sites, as illustrated in **Figure 19**.

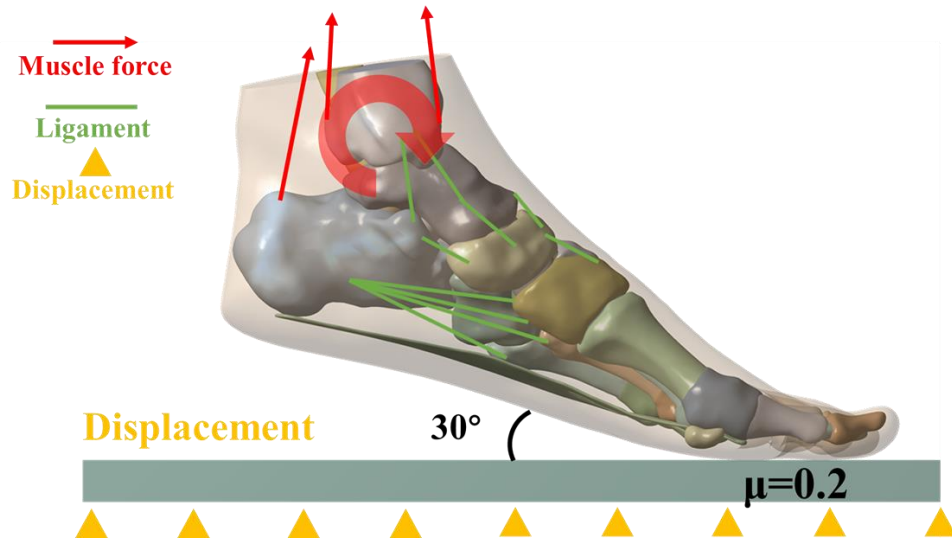


Figure 19. Loading and boundary conditions applied to the foot model

The ground, modeled as a rigid plate, was constrained to allow movement solely in the vertical direction. A displacement-controlled iterative loading approach was employed,

wherein the vertical displacement of the ground plate was iteratively adjusted until the reaction force predicted by the simulation converged with the magnitude of the experimentally measured Vertical Ground Reaction Force (vGRF) [117]. The specific loading conditions are showed in **Table 1**.

2.6.6 Knee Boundary Conditions and Loading during Stair Ascent

For the single-leg stance phase of stair ascent, the contact and boundary conditions of the model were defined as follows: The contact between the menisci and the femoral/tibial cartilage was defined as Frictionless contact, allowing relative sliding between surfaces. Conversely, the interface between the articular cartilage and the underlying subchondral bone was defined as Bonded contact to simulate their tight anatomical connection [93].

A 6-DOF full constraint was applied to the distal cross-sections of the tibia and fibula, restricting translation and rotation in all directions. The calculated average muscle forces of the Quadriceps, Biceps Femoris, and Gastrocnemius were applied to their corresponding locations on the model. Additionally, a vertical downward concentrated force was applied to the proximal femur to simulate the compressive effect of Body Weight on the knee joint, as illustrated in **Figure 20**.

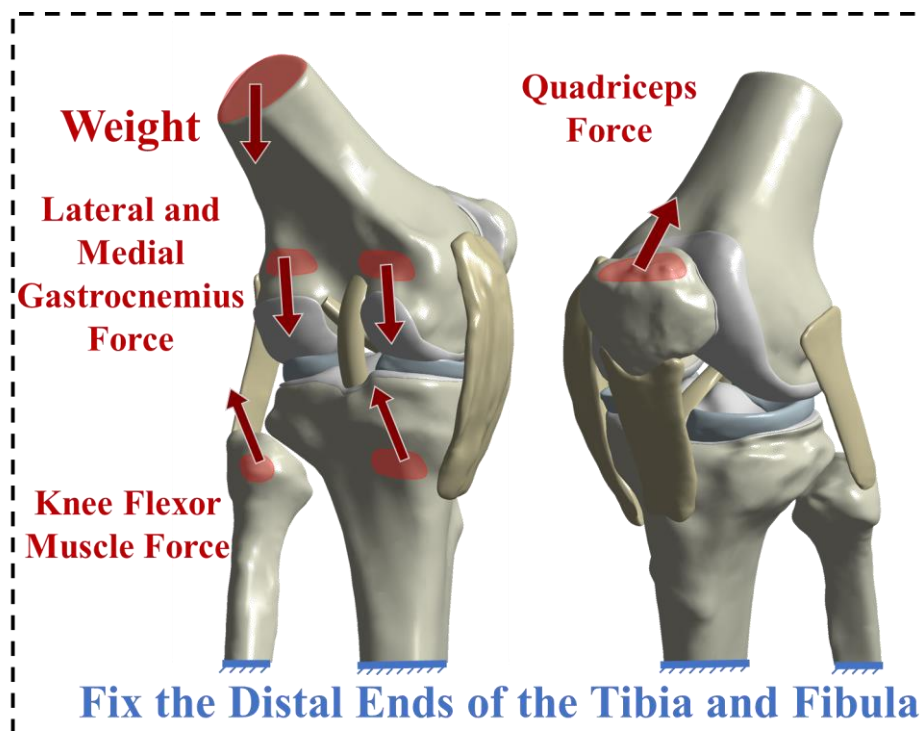


Figure 20. Loading and boundary conditions of the knee joint during stair ascent

2.6.7 Knee Boundary Conditions and Loading during Jump Landing

To address the high-impact jump landing maneuver, six Transient Analyses were conducted to evaluate the in vivo biomechanical status of the PT during the landing phase. Consistent with the stair ascent model, the contact interface between the menisci and cartilage was defined as frictionless, while the interface between the cartilage and bone was defined as bonded contact.

A 6-DOF full constraint was applied to the distal ends of the tibia and fibula. In contrast to the quasi-static loading employed for stair ascent, the landing model utilized more complex dynamic loading. The experimentally acquired 3D Knee Joint Moments and Quadriceps muscle forces (averaged from 10 datasets) were applied as time-dependent dynamic loads to the distal femur and the quadriceps tendon, respectively. The specific implementation is illustrated in **Figure 21**.

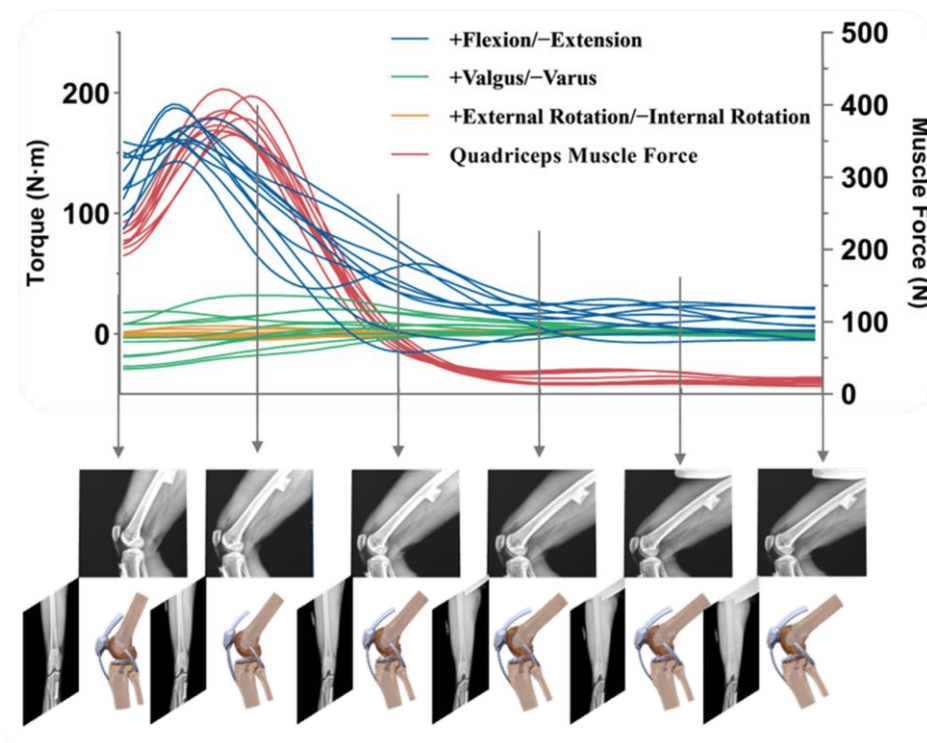


Figure 21. Simulated knee joint loads during jump landing

2.7 Image Acquisition and Registration

2.7.1 Acquisition of Dynamic Foot Imaging Data

This study utilized the DFIS to validate the effectiveness of the foot FE model. The specific focus was on measuring dynamic changes in the length and height of the MLA, as well as the width of the Rearfoot Transverse Arch during locomotion.

The DFIS primarily consists of two paired X-ray sources and Image Intensifiers. The source-to-receiver distance was set to 132 cm. The two image intensifiers were positioned in a 90° orthogonal configuration to simultaneously acquire fluoroscopic images of the foot in both the sagittal and coronal planes.

The subject walked on a treadmill at a speed of 1.4 m/s (approximately 5 km/h). During data acquisition, the tube voltage was set to 65 kVp and the tube current to 0.80 mA. With a pulsed fluoroscopy duration of 20 seconds, the average radiation dose received by the subject was approximately 106 mrem. Images were stored in 8-bit grayscale format with a resolution of 1024×1024 pixels, corresponding to an effective field of view of 315mm. Additionally, images of a perforated plate for distortion correction and calibration beads for reconstructing fluoroscopic geometric parameters were acquired before and after the experiment. The collected dynamic images are illustrated in **Figure 22**.

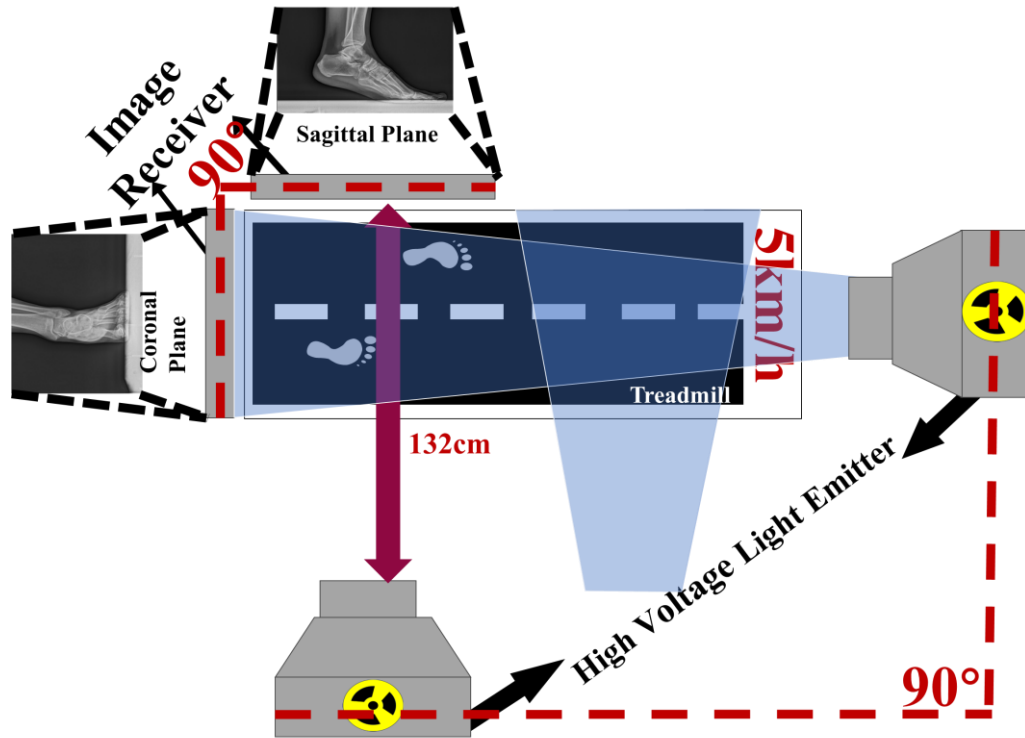


Figure 22. Dynamic images of the foot captured by the Dual-plane Fluoroscopy System

2.7.2 Acquisition of Dynamic Knee Imaging Data during Stair Ascent

A DFIS was employed to acquire dynamic X-ray images of the knee joint at a frame rate of 60 Hz. The system comprises two sets of orthogonally positioned X-ray sources and image intensifiers. To achieve the optimal field of view, the staircase apparatus was positioned at a 45° angle relative to the two image intensifiers, as shown in **Figure 23**. X-ray images corresponding to five key instances—0%, 25%, 50%, 75%, and 100% of the right leg Stance Phase were extracted for subsequent registration [118].

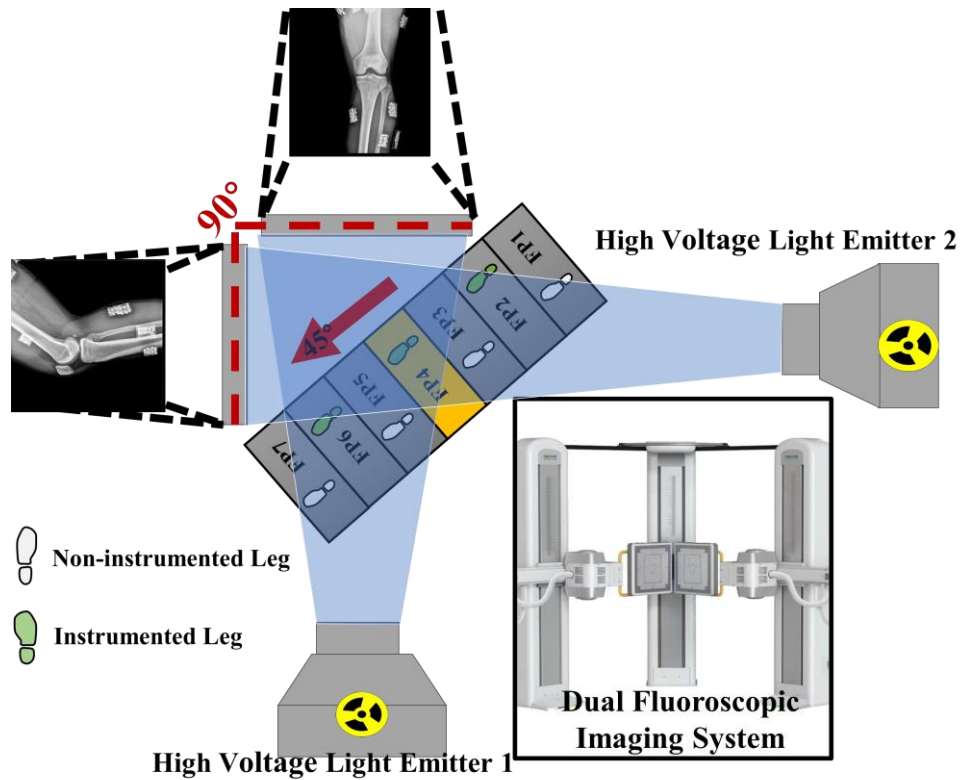


Figure 23. Acquisition of dynamic knee joint images during stair ascent.

2.7.3 Acquisition of Dynamic Knee Imaging Data during Jump Landing

Consistent with the knee stair ascent experiment described previously, a high-speed DFIS system was employed to capture dynamic images of the jump landing phase (spanning from Initial Contact to Maximum Knee Flexion) at a frequency of 60 Hz. Prior to the trials, the system was strictly calibrated according to established protocols, including the determination of the imaging field of view, spatial geometric calibration, and image distortion correction. Key frame images corresponding to the landing phase were automatically extracted by the analysis system for subsequent processing.

2.8 Knee Joint Registration

A virtual DFIS environment was built in Solidworks, where anatomical landmarks were identified. The 3D bone model was then aligned to best fit the projection outline from DFIS images.

The bones performed flexion/extension around the mediolateral axis, varus/valgus rotation around the anteroposterior axis, and internal/external rotation around the

proximodistal axis. To analyze the tibiofemoral joint's six degrees of freedom (6DOF) kinematics, coordinate systems were defined based on previous studies[119, 120]. Tibial local coordinate system $X_tY_tZ_t$. O_t is the origin, defined as the midpoint of the line connecting the lateral and medial tibial plateaus; X_t is the line connecting the lateral and medial tibial plateaus and is directed toward the lateral plateau; Y_t is perpendicular to both X_t and Z_t . Femoral local coordinate system $X_fY_fZ_f$. O_f is the origin, defined as the midpoint of the line connecting the lateral and medial femoral condyles; X_f is the line connecting the lateral and medial femoral condyles and is directed toward the lateral condyle; Y_f is perpendicular to both X_f and Z_f , as shown in **Figure 24**.

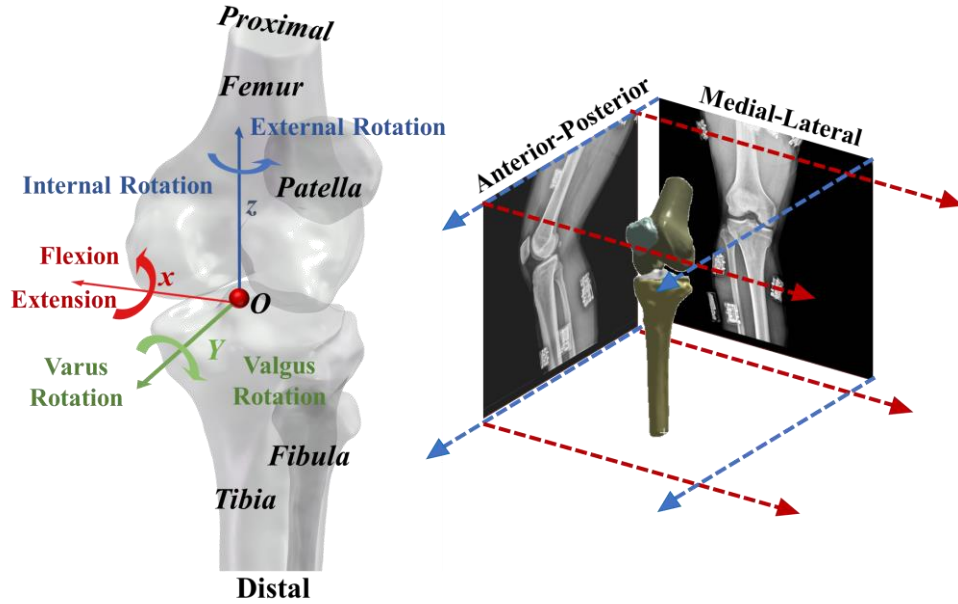


Figure 24. Knee Joint Registration

Using the defined coordinate systems, 6DOF tibiofemoral joint kinematics were computed, with in vivo knee motion represented as the movement of the tibial system relative to the femoral system. Rotational transformation between the tibial and femoral local coordinate systems. From Eq. (3), the femoral local coordinate system $X_fY_fZ_f$ is transformed by the rotation R to obtain the tibial local coordinate system $X_tY_tZ_t$. By computing α, β, γ from Eq. (4), the kinematic parameters of the femur's rotation relative to the tibia are obtained.

$$R: F \rightarrow T \quad (3)$$

$$R = \begin{bmatrix} \cos\gamma & \sin\gamma & 0 \\ -\sin\gamma & \cos\gamma & 0 \\ 0 & 0 & 1 \end{bmatrix} \cdot \begin{bmatrix} \cos\beta & 0 & -\sin\beta \\ 0 & 1 & 0 \\ \sin\beta & 0 & \cos\beta \end{bmatrix} \cdot \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos\alpha & \sin\alpha \\ 0 & -\sin\alpha & \cos\alpha \end{bmatrix} = \begin{bmatrix} R_{11} & R_{12} & R_{13} \\ R_{21} & R_{22} & R_{23} \\ R_{31} & R_{32} & R_{33} \end{bmatrix} \quad (4)$$

Eq. (4) corresponds to a Z-Y-X rotation sequence $R = R_z(\gamma)R_y(\beta)R_x(\alpha)$. Thus, $\beta = \arcsin(R_{31})$, $\alpha = \text{atan2}(R_{32}, R_{33})$, and $\gamma = \text{atan2}(-R_{21}, R_{11})$. In the gimbal-lock case ($|\cos\beta| \approx 0$), $\beta = \pm\pi/2$ and the remaining angle is obtained from $\text{atan2}(R_{12}, R_{22})$ with a continuity constraint.

2.9 Validation of FE Models

2.9.1 Validation of the Foot FE Model

Because the flexion angle of MTPJ simulated by FE method is 8.62 ° and the starting angle is 30 °, the length and height of MLA of foot and the average and standard deviation of length and height of posterior transverse arch of foot are measured when subjects walk at 5m/s speed and the dorsiflexion angle of MTPJ of 10 gait cycles is 38.62 °. The distance between the medial tubercle of calcaneus and the base of the first metatarsal head was measured, that is, the length of the longitudinal arch of the foot, the height of the longitudinal arch of the foot from the scaphoid bone to the longitudinal arch of the foot [79], the width of the posterior transverse arch of the foot from the base of the first metatarsal bone to the base of the fifth metatarsal bone and the distance from the base of the second metatarsal bone to the first metatarsal bone to the fifth metatarsal bone. In order to understand the change of the windlass mechanism in the later stance phase of the gait cycle, the flexion angle of the first MTPJ on the sagittal plane, as shown in **Figure 25**, the MLA angle, and the transverse arch angle of the hindfoot on the coronal plane were measured [121].

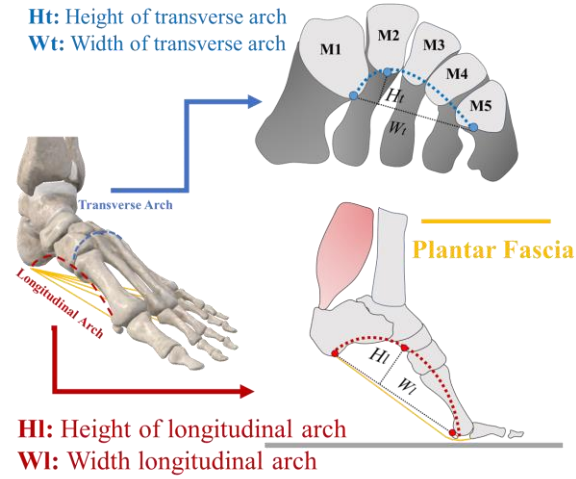


Figure 25. Measurement of transverse and longitudinal arch parameters.

2.9.2 Validation of the Knee FE Model during Stair Ascent

The validity of the model was verified by comparing ligament length variation data from in vivo biomechanical studies. Based on the flexion angles of 25 knee joint models, the strain parameters of the ACL and PCL were extracted within the flexion range of 0° – 60° , and validated against in vivo experimental data reported by Li et al. [122].

2.9.3 Validation of the Knee FE Model during Jump Landing

Pearson correlation analysis was conducted to assess the consistency between muscle activation results obtained from electromyography (EMG) sensors and those simulated from OpenSim. Square of the correlation coefficient (R^2) greater than 0.49 indicates strong correlation[123].

To validate the effectiveness of the FE model, we applied a 134 N anterior force to the tibia in a fully extended knee position (0° flexion) while fixing the femur and allowing the tibia to move only in the anterior-posterior direction [124, 125]. The degree of anterior tibial translation was then compared with results obtained from similar experiments conducted by other researchers.

To establish a predictive relationship between easily measurable external biomechanical loads and the internal stress state of the PT, we employed a Ridge Regression analysis.

This method was selected specifically to address multicollinearity, a common issue in biomechanical data where predictor variables are highly correlated. Unlike ordinary least squares OLS regression, which can produce unstable estimates with high variance when predictors are correlated, Ridge Regression introduces a regularization penalty to ensure model stability and generalizability [126].

The dataset consisted of $N = 60$ samples, derived from the six distinct time frames of the landing phase across ten repeated stop-jump trials. The dependent variable (y) was defined as the peak equivalent (von Mises) stress within the PT obtained from the FE simulation.

The regression model is formulated as Eq(5):

$$y = \beta_0 + \beta_1 x_1 + \beta_2 x_2 + \cdots + \beta_n x_n + \epsilon \quad (5)$$

The independent variables (predictors, x) included four kinetic parameters: Knee flexion moment (x_1), Internal-external rotation moment (x_2), Varus-valgus moment (x_3), Quadriceps muscle force (x_4). Prior to model fitting, all predictor variables were standardized (z-score normalization) to have a mean of 0 and a standard deviation of 1. This step is critical in Ridge Regression to ensure that the regularization penalty is applied uniformly across all variables, preventing predictors with larger physical magnitudes from dominating the model artificially.

The Ridge Regression model estimates the coefficients (β) by minimizing the residual sum of squares extended by a penalty term proportional to the square of the magnitude of the coefficients (L2 regularization). The objective function is formulated as Eq(6):

$$\hat{\beta} = \underset{\beta}{\operatorname{argmin}} \left(\sum_{i=1}^n \left(y_i - \beta_0 - \sum_{j=1}^p x_{ij} \beta_j \right)^2 + \lambda \sum_{j=1}^p \beta_j^2 \right) \quad (6)$$

Where: y_i is the observed maximum equivalent stress for the i -th sample ($i = 1, \dots, n$, where $n = 60$), x_{ij} is the value of the j -th predictor variable for the i -th sample

($j = 1, \dots, p$, where $p = 4$), β_0 is the intercept term, β_j are the regression coefficients for the j -th predictor, λ is the complexity parameter (tuning hyperparameter) that controls the strength of the shrinkage penalty; a larger λ implies greater shrinkage to reduce model complexity and prevent overfitting.

The dataset was randomly partitioned into a training set (80%) for model fitting and a testing set (20%) for independent evaluation. To determine the optimal value of λ , we utilized Leave-One-Out Cross-Validation (LOOCV) on the training set. The λ value that minimized the mean squared error (MSE) during cross-validation was selected for the final model.

The predictive performance of the model was evaluated using the Coefficient of Determination (R^2) and the Adjusted R^2 to penalize for model complexity. The statistical significance of the regression was assessed using an F-test ($p < 0.001$). Residual analysis was conducted to verify assumptions of linearity and homoscedasticity, ensuring the error terms (ϵ) were normally distributed and independent.

3 Results

3.1 Experimental and Simulation of Foot Biomechanics

3.1.1 Validity of FE Model

The width of transverse arch, height of longitudinal arch, length of longitudinal arch simulated by the PF model with 100% stiffness in this study was compared with the data measured by DFIS. Because of the visual deviation of the height of transverse arch with the increase of the dorsiflexion angle of the MTPJ, the height of transverse arch was not recorded. Results as shown in **Figure 26**, the width of transverse arch, height of longitudinal arch, length of longitudinal arch measured by DFIS at 38.62 °dorsiflexion of MTPJ is $45.14 \pm 1.63\text{mm}$, $29.29 \pm 1.57\text{mm}$, $155.16 \pm 2.69\text{mm}$, respectively. The results of FE simulation are 46.51mm, 29.96mm, 156.71mm, respectively. The simulation results are similar to the experimental results, so it is proved that the model can accurately predict the changes of foot arch during walking.

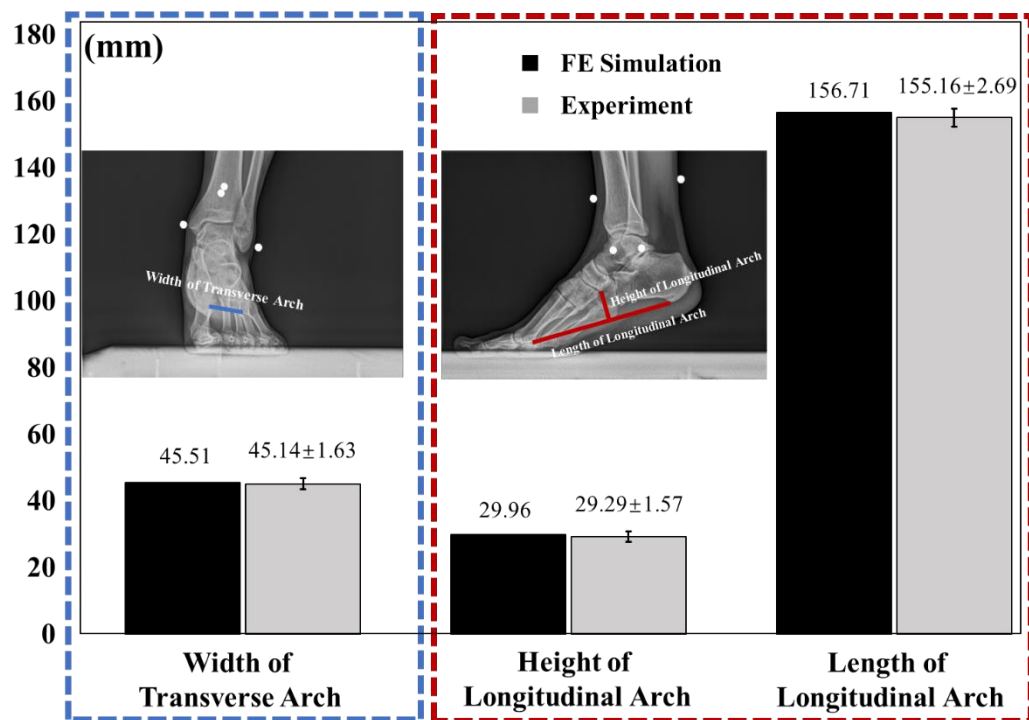


Figure 26. The arch data measured by DFIS and the data simulated by FE method

3.1.2 Longitudinal Arch and MTPJ

As shown in **Figure 27 A-C**, the length, height and MLA change with the change of the stiffness of the PF. The simulation results of **figure 27A** show that the length of the longitudinal arch of the foot decreases with the increase of the stiffness of the PF. As shown in **figure 27A**, the length of the longitudinal arch increases by 2.06mm, 1.21mm, 0.50mm, respectively, when the PF is compared with the stiffness of the PF (156.71mm). When the PF was 125%, 150%, 175%, 200%, the length of the longitudinal arch was shortened by 1.19mm, 2.8mm, 4.51mm.

Figure 27B shows the change of the height of the longitudinal arch with the stiffness of the PF. With the increase of the stiffness of the PF, the height of the longitudinal arch increases. Compared with 29.96mm, the height of the longitudinal arch decreases by 1.35mm, 0.92 mm, 0.72 mm, respectively, when the PF is at 25% 50% 75% stiffness. The PF was increased by 0.05mm, 0.18mm, 0.28mm, 0.66mm under 125%, 150%, 175%, 200% stiffness, respectively.

Figure 27C shows the variation of the MLA with the stiffness of the PF. The simulation results show that the MLA decreases with the increase of the PF stiffness. The PF increases 135.82 °, 2.79 ° and 1.72 ° respectively under the stiffness of 25% 50% 75 °. The PF decreased the MLA by 0.96 °, 3.25 °, 6.42 ° and 7.38 respectively when the stiffness of PF was 125% 150% 175% 200%.

Figure 27D shows the change of the dorsiflexion angle of the first MTPJ with the PF stiffness, and the dorsiflexion angle of the first MTPJ decreases with the increase of PF stiffness.

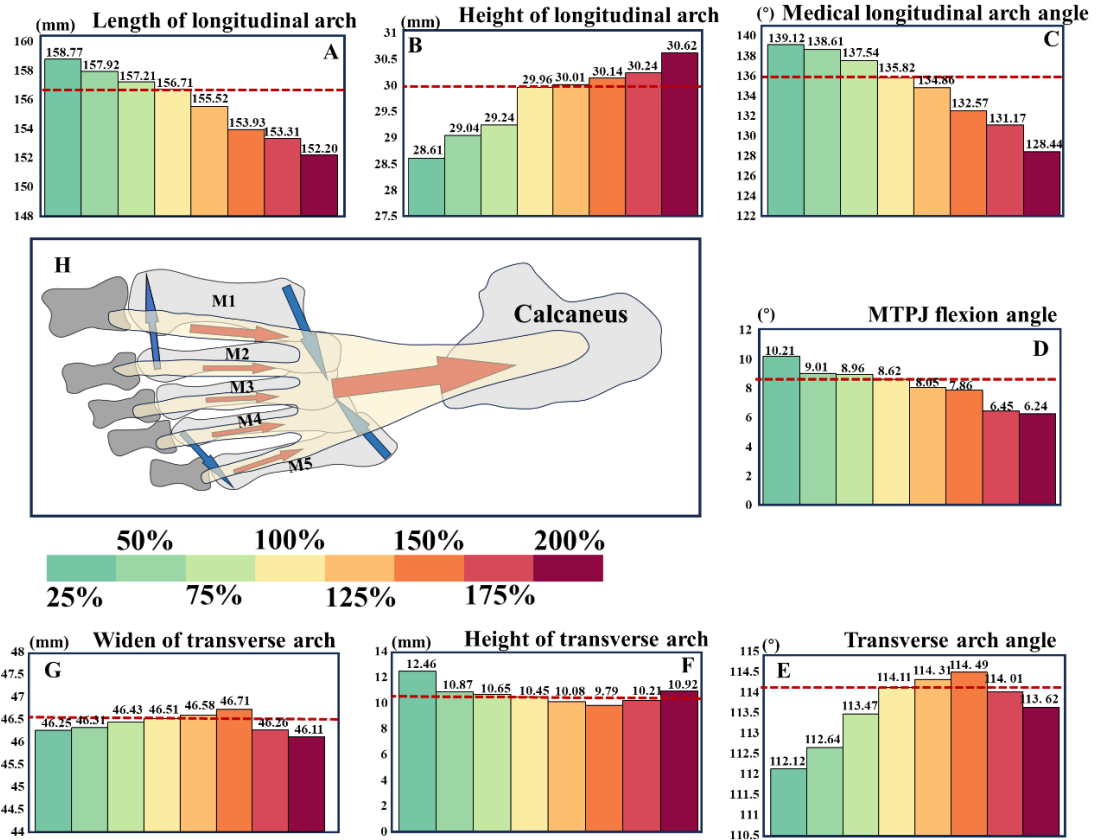


Figure 27. A-G: The changes of foot arch and MTPJ flexion angle under different PF stiffness; H: The tension on the distal end of the PF spreads the distal end of the metatarsal outward and tightens the proximal end inward

3.1.3 Transverse Arch

Figure 27G-E shows the changes of the width, height, and TA of the foot respectively. As shown in **Figure 27G**, when the stiffness of the PF is 25%-150%, the width of the transverse arch increases with the increase of the stiffness of the PF, while when the stiffness of the PF is 175%-200%, the width of the transverse arch decreases.

Figure 27F shows the change of the foot transverse arch height. When the PF stiffness is 25%-150%, the foot transverse arch height decreases with the increase of the PF stiffness, while when the PF stiffness is 175%-200%, the foot transverse arch height decreases with the increase of the PF.

Figure 27E shows the change of the TA. When the PF stiffness is between 25% and 150%, the TA increases with the increase of the PF stiffness, and the transverse arch

stiffness decreases with the increase of the PF stiffness. When the PF stiffness is 175%-200%, the TA decreases, and the transverse arch stiffness increases with the increase of PF stiffness.

3.1.4 PF Stress

From the stress distribution of the PF (**Figure 28A**), the stress of the PF is mainly concentrated in the fascia bundle below the first MTPJ, and on the inside of the PF. With the increase of the stiffness of the PF, the stress on the inside of the PF is more concentrated. From the stress value (**Figure 28B**), with the increase of the stiffness of the PF, the stress in the middle and distal part of the PF increases. The stress at the proximal end of PF showed a downward trend.

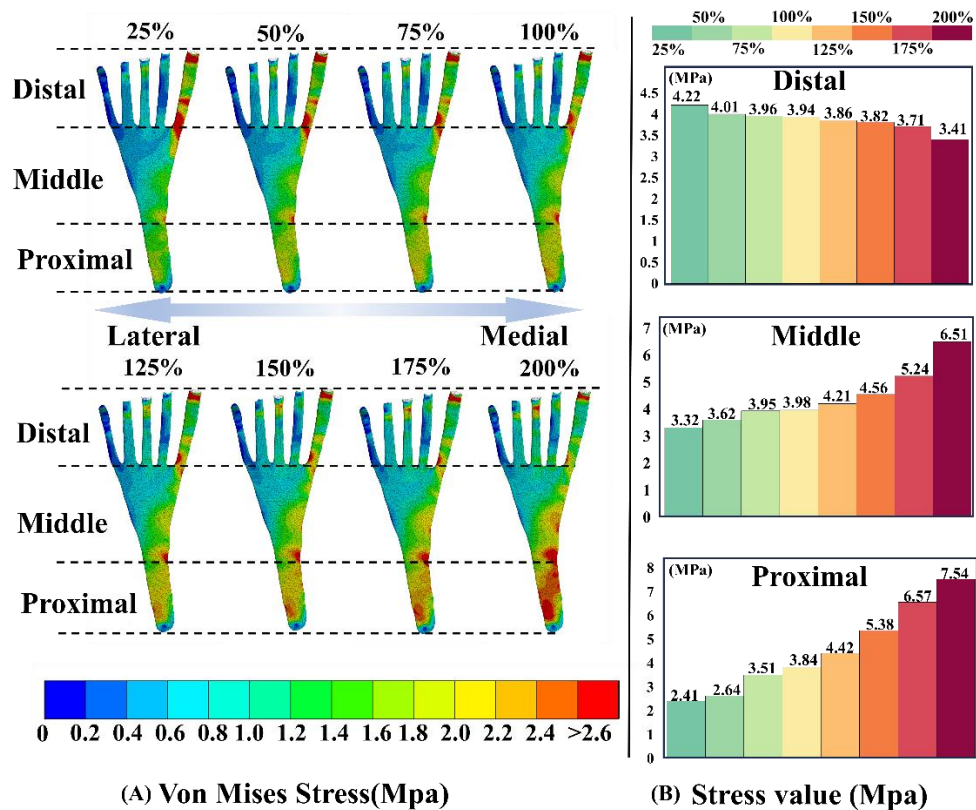


Figure 28. Changes of stress distribution and maximum stress value of PF with different stiffness

3.2 Biomechanical Experimental and Simulation of Stair Ascent

3.2.1 Knee Joint Angle and Muscle Force

Based on the dynamic X-ray images of the right-leg support phase during the 120-200 mm stair test, this study performed registration analysis of the 3D skeletal model. The results showed that for each tested stair height, five motion phase models could be reconstructed, resulting in a total of 25 sets of 3D skeletal motion models. Measurements revealed that the range of flexion angles of the femur relative to the tibia varied from 3.24° to 62.41° , as shown in **Figure 29**, with flexion angles increasing as the stair height increased. At the same time, the support phase was accompanied by both knee varus and external rotation.

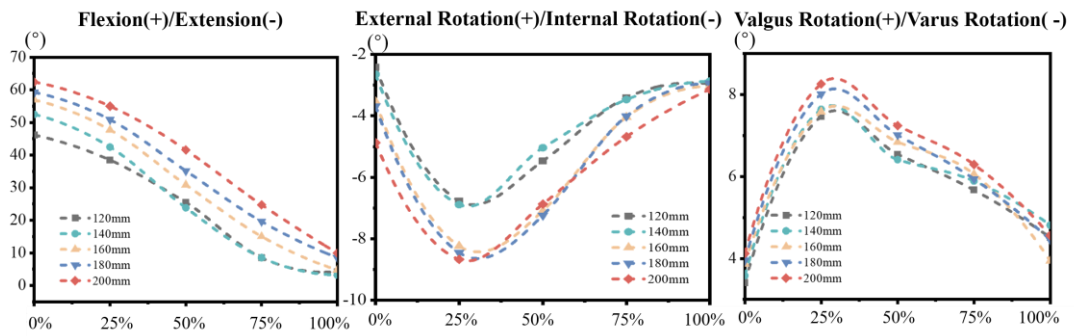


Figure 29. Angle of Femoral Relative Tibial Motion

The muscle activation levels of the subjects during stair climbing, calculated using the OpenSim static optimization algorithm, were compared with the EMG signals recorded by the Delsys EMG testing system during the experiment. The comparison shows similar trends in the muscle activity signals, as shown in the **Figure 30a**, indicating that the data from the OpenSim musculoskeletal model is reliable to some extent in this study [127, 128]. Finally, muscle forces at the 0%, 25%, 50%, 75%, and 100% time points were selected as loading conditions and imported into the FE model, as shown in the **Figure 30b**.

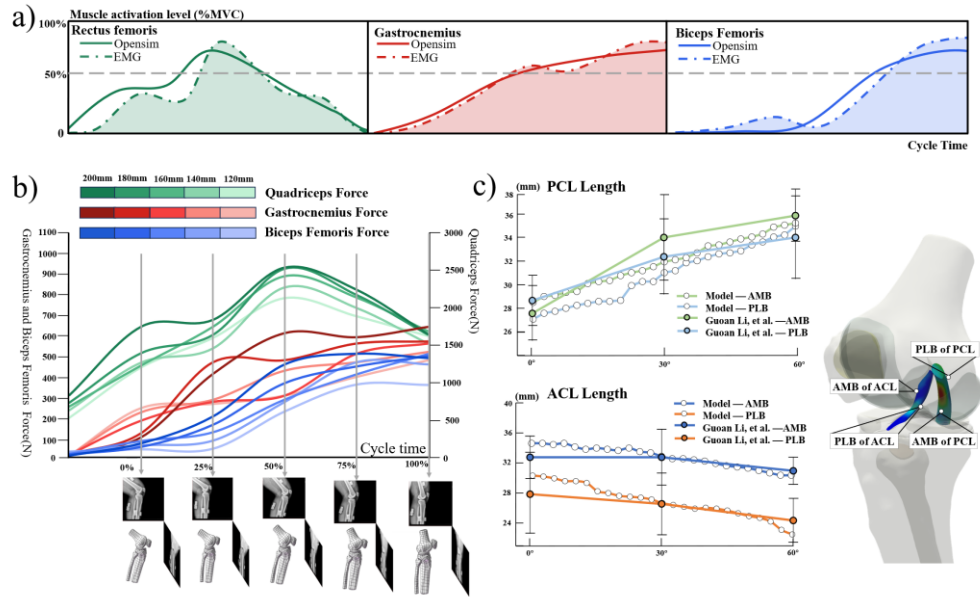


Figure 30. a) Comparison between the muscle activation levels calculated by OpenSim and the EMG signal activity recorded by the Delsys EMG testing system. b) Muscle forces at different stair heights. c) Comparison between the established model and the in vivo experimental data from Li et al.

3.2.2 The validity of the model

The ACL and PCL are divided into the anterior-lateral bundle (ALB) and posterior-medial bundle (PMB). According to Li et al. [122], during knee flexion from 0° to 60°, the ALB of ACL shortened by 5.54%, the PMB shortened by 12.68%, the ALB of PCL length increased by 27.34%, and the PMB length increased by 17.01%. In this knee joint model, with a flexion angle ranging from 3.24° to 62.41°, the ALB of ACL shortened by 8.74%, the PMB shortened by 16.82%, the ALB of PCL increased by 24.74%, and the PMB increased by 16.78%. Comparing the ligament length change rates calculated by the FE model with the literature data, the ligament length change rates at different flexion angles in this study are generally consistent with the trends reported in the literature, as shown in **Figure 30c**. This verifies the accuracy of the model's kinematics and provides a reliable foundation for the subsequent biomechanical study of the knee joint.

3.2.3 Meniscus Stress

This study divided the meniscus into six regions: the anterior horn (MA), body (MB), and posterior horn (MP) of the medial meniscus, and the anterior horn (LA), body (LB), and posterior horn (LP) of the lateral meniscus. With increasing stair height, overall meniscal stress showed an upward trend. Stress concentration was most pronounced between 25%–75% of the cycle, the transition from leg lift to foot contact, primarily in the body regions. At a stair height of 200 mm, stresses were higher and more widely distributed, with larger high-stress areas, as shown in **Figure 31**.

In the medial meniscus, stresses were mainly borne by the body and posterior horn. MP showed relatively high stresses from 0% to 25% of the stance phase, with its peak increasing from 2.47–2.83 MPa at 120–140 mm to 4.31 MPa at 200 mm, indicating marked height sensitivity in early–mid stance. MB exhibited the highest overall stresses, with clear amplification at greater stair heights; its maximum reached 5.61 MPa at 50% of the stance phase for the 200 mm stair, exceeding the values in MA and MP. MA stresses were negligible at 0–25% but increased in late stance, approaching 2.0–2.3 MPa at 75–100% for all heights, as shown in **Figure 31**.

In the lateral meniscus, overall stresses were lower than on the medial side, but a similar temporal redistribution was observed. LB carried the largest share of lateral loading, with peak values around 3.0 MPa at higher stair heights and later time points. LA remained lightly loaded and only became appreciable after 50% of the stance phase, whereas LP showed moderate stresses with a peak of 2.29 MPa at 50% for the 200 mm stair. These patterns indicate that increasing stair height primarily augments loading in the medial body and posterior horn while moderately elevating stresses in the lateral body and posterior horn, as shown in **Figure 31**.

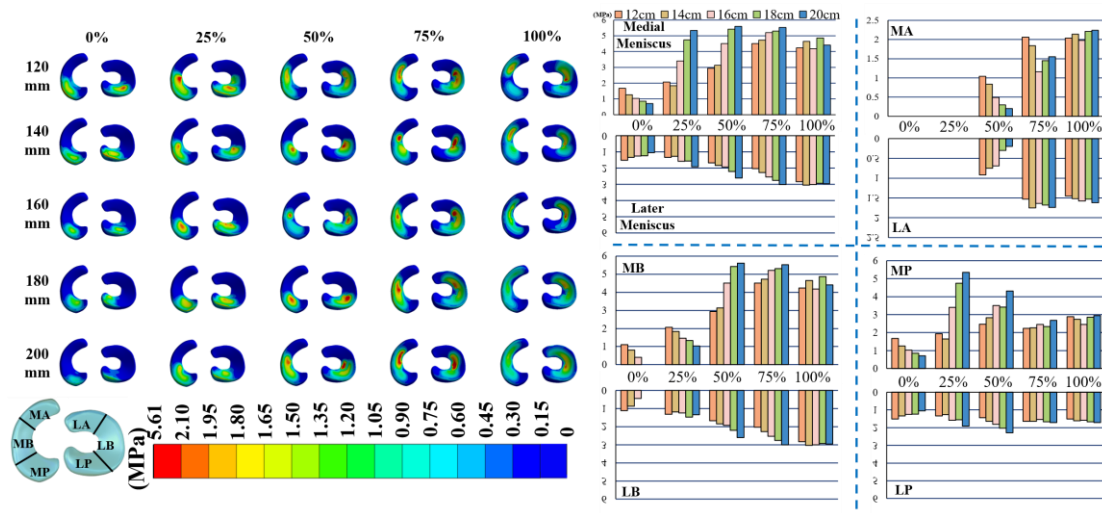


Figure 31. Stress distribution of the medial and lateral menisci and variation in maximum stress values in the anterior horn, body, and posterior horn of the menisci.

3.2.4 ACL/PCL stress

Under different stair heights, the PCL stress–time curves all exhibited a pattern of initial increase followed by a decrease over the single-leg support phase as shown in **Figure 32**. At 25% of the support phase, the PCL stress reached its primary peak, with values of 0.65, 0.74, 0.87, 1.20, and 1.32 MPa for stair heights of 120, 140, 160, 180, and 200 mm, respectively, showing an approximately monotonic increase with increasing stair height. Compared with the peak value of 0.65 MPa at 120 mm, the peak at 200 mm increased to 1.32 MPa. At 0%, 50%, 75%, and 100% of the support phase, PCL stress generally remained within the range of 0.24–0.62 MPa, and the increase induced by stair height was clearly smaller than that observed at the 25% peak, indicating that stair height primarily amplifies PCL stress in the early support phase, as shown in **Figure 32**.

The ACL stress–time curves showed a typical single-peaked pattern: stress was low in the early support phase, then rose rapidly over time and reached its peak at 50% of the support phase. At this time point, ACL stress for stair heights from 120 mm to 200 mm was 0.62, 0.64, 0.75, 0.80, and 0.84 MPa, respectively, with the peak generally increasing with stair height. At 0%, 25%, 75%, and 100% of the support phase, ACL stress ranged from 0.06 to 0.64 MPa; although some fluctuations with stair height were observed, the

magnitude of these changes was smaller than at the peak time point. Overall, increasing stair height elevates the peak stresses of both the PCL and ACL, with a more pronounced effect on PCL stress in the early support phase, whereas the ACL bears the major load in the mid-support phase, as shown in **Figure 32**.

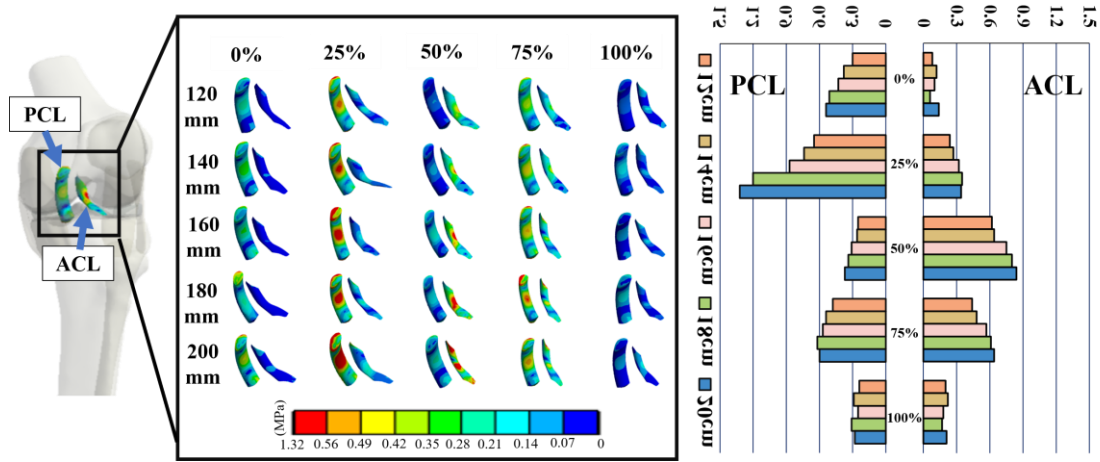


Figure 32. Stress distribution and stress values of the ACL and PCL

3.3 Experimental and Simulation of Jump Landing

3.3.1 Model Validation

Figure 33A presents the comparison between muscle activation levels obtained from EMG sensors and those predicted by the musculoskeletal model using OpenSim for the vastus lateralis, rectus femoris, and vastus medialis muscles. The results show a strong correlation between the simulated and experimental data, with R^2 values of 0.8880, 0.7129, and 0.8868, respectively. This high degree of correlation suggests that the musculoskeletal model employed in this study provides reliable and accurate predictions of muscle activation patterns.

The FE model was validated by applying a 134 N anterior force to the tibia and measuring the resulting anterior tibial translation, which was found to be 4.81 mm. The anterior tibial translation data were compared with previous studies under similar loading conditions. The agreement between our model's predictions and the results of previous studies as shown in **Figure 32B**, confirms the validity of the model in this study.

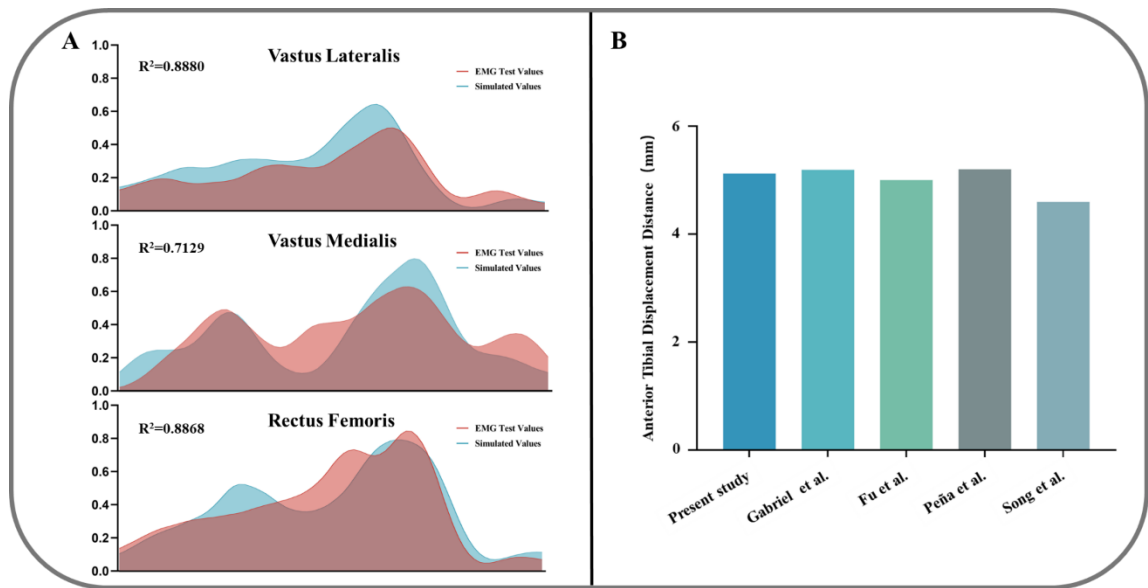


Figure 33. Result of model validation (A) Fitting results of muscle forces simulated by OpenSim and EMG signals for the vastus lateralis, rectus femoris, and vastus medialis. Red indicates simulated values, and blue indicates measured values. (B) Validation of the FE model against results from other studies.

3.3.2 Maximum stress and stress distribution in the PT

Figure 34 shows the FEA results of the PT during the landing phase. The maximum equivalent stress was highest at the tendon's attachment sites to the bone at all six analyzed time points. At initial ground contact (Moment 1), the maximum stress reached 94.44 MPa. This stress progressively decreased to 88.88 MPa (Moment 2), 27.41 MPa (Moment 3), 26.68 MPa (Moment 4), 16.37 MPa (Moment 5), and finally, 21.62 MPa at Moment 6, with stress concentration shifting from the tibial attachment to the patellar attachment.

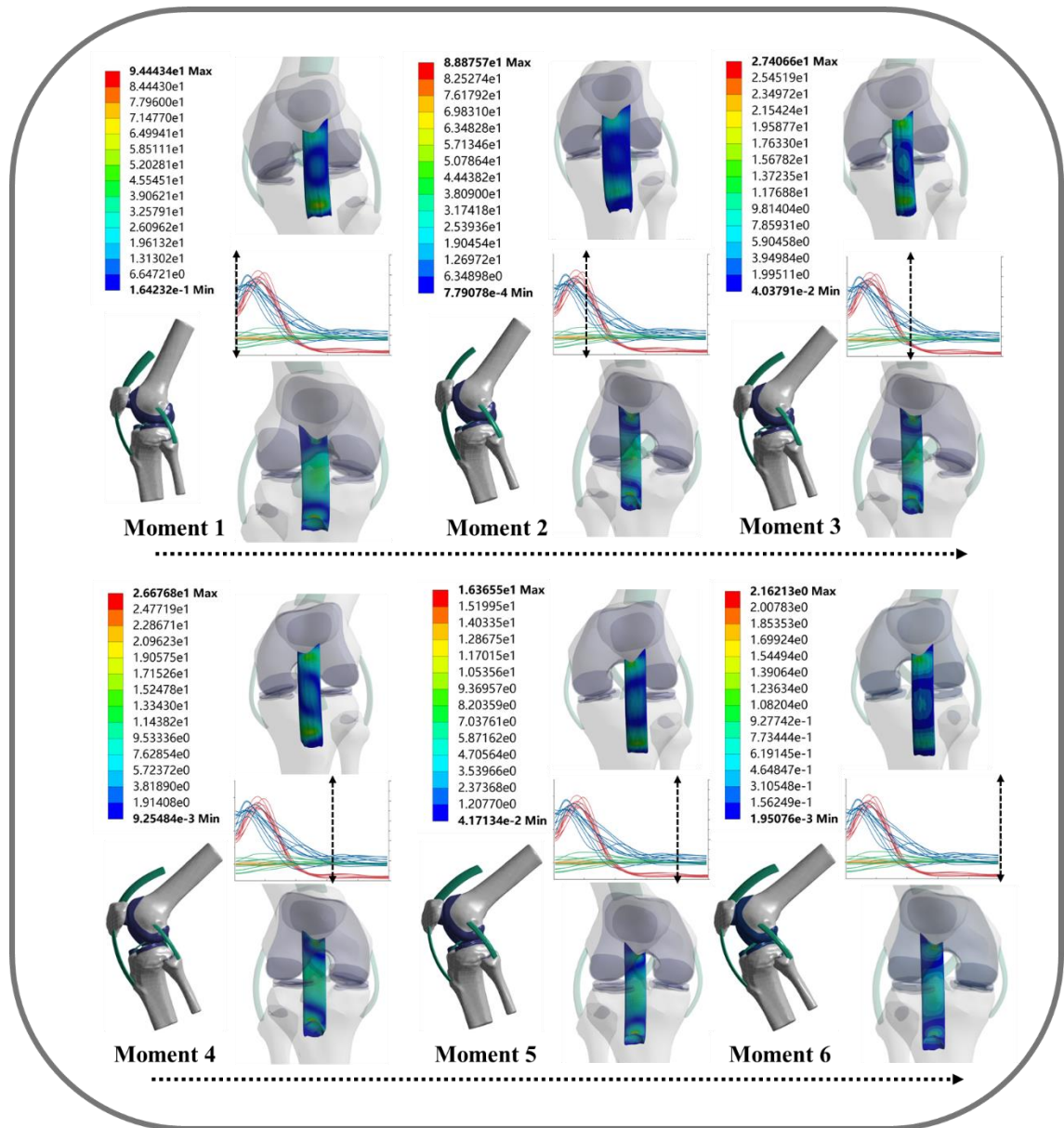


Figure 34. FEA of the PT during landing phase. Includes the equivalent (von-Mises) stress simulation results at six different moments during the landing phase.

3.3.3 Results of PT maximum equivalent (von-mises) stress prediction

Figure 35 shows the results of our ridge regression analysis. The central scatterplot compares simulated versus predicted stresses for the training and test sets, the vast majority of points lie close to the line without systematic bias. The upper-left histogram shows that residuals are symmetrically distributed around zero and confined within ± 5 MPa. The upper-center and right-side histograms display the distributions of simulated

and predicted stresses, respectively, for both sets; these largely overlap, confirming consistent model performance across datasets with no indication of overfitting or distributional shift. Statistically, the model demonstrated strong explanatory power ($R^2 = 0.859$, adjusted $R^2 = 0.851$) and is highly statistically significant ($F = 160.758$, $p < 0.001$). Among the predictors, knee flexion moment has the strongest effect ($\beta = 0.747$, $t = 17.461$, $p < 0.01$), followed by quadriceps muscle force ($\beta = 0.181$, $t = 6.248$, $p < 0.01$), whereas internal-external rotation ($\beta = 0.014$, $t = 0.417$, $p = 0.679$) and varus-valgus moment ($\beta = 0.084$, $t = 0.249$, $p = 0.327$) contribute negligibly.

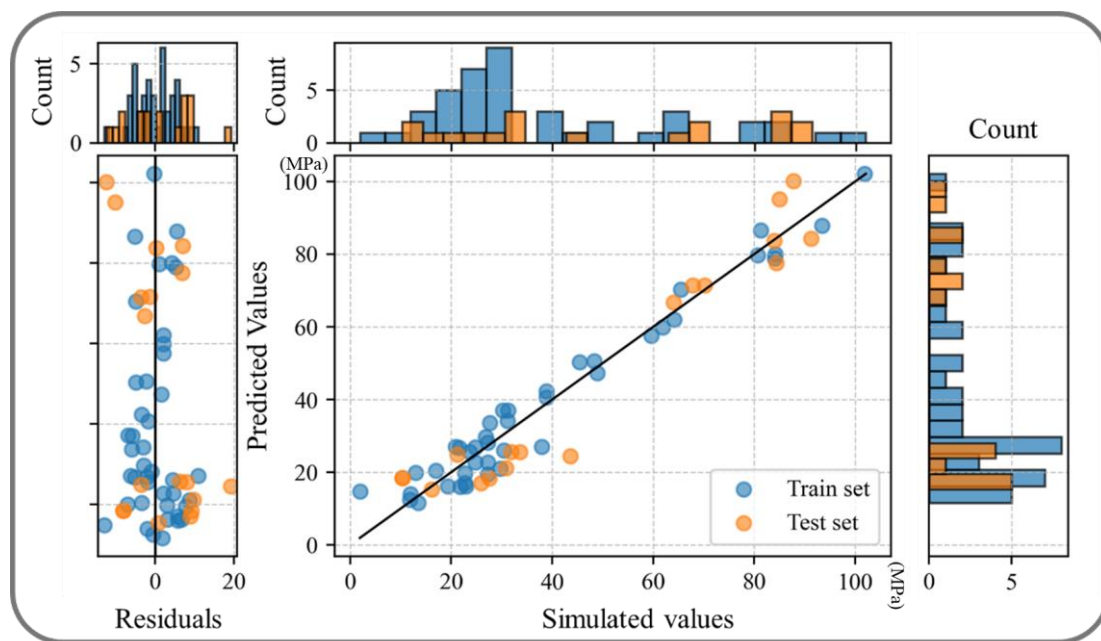


Figure 35. Ridge Regression Analysis. Left: Residual analysis—top, residual histograms (training in blue, testing in orange); bottom, scatter plot of residuals versus predicted values with a zero-reference line. Center: Simulated data—top, histograms; bottom, simulated versus predicted scatter plot with an identity line. Right: Predicted values—horizontal histograms for both datasets, sharing the y-axis.

4.Disscion

4.1 Effect of PF Stiffness on the Windlass Mechanism

DFIS is a new type of in vivo joint imaging technique, which is non-invasive, highly compatible, dynamically captures bone motion in the body, and is not affected by the relative motion of skin and other soft tissues. It is the latest motion analysis method for non-invasive and accurate analysis of foot and ankle motion and evaluation of joint function. In the technical aspect, based on the dynamic X-ray imaging technology, DFIS can directly dialyze the bone position and motion law of different shapes and complex structures in the foot and ankle, which can make up for the limitation that the traditional biomechanical measurement methods cannot see the real motion of the joint. In terms of accuracy, the accuracy of DFIS in determining the joint position and capturing joint motion is up to sub-millimeter level [24]. The in vivo six degrees of freedom of constant bony structure can be accurately captured from the point of view of the real in vivo motion of the joint, quantifying the variation characteristics of the lengths of the longitudinal arch and the transverse arch during walking. The FE simulation results are compared with the data measured by DFIS. The errors of the longitudinal arch length and the longitudinal arch width higher than the transverse arch width are 3.03%, 2.29%, 0.99%, respectively. It further illustrates the effectiveness of the FE method in predicting the change of arch length.

The windlass mechanism effect of PF is very important to maintain foot stiffness in human gait. From the middle of the stance phase to the end of the brace, the flexion of the MTPJ can increase the tension of the PF, and then shorten the longitudinal arch by flexion and adduction of the MTPJ with supination of the posterior foot. The changes between these bone structures strengthen the stiffness of the foot, attenuate the cushioning performance of the foot, and make the foot transition into a rigid lever, which can effectively transfer the metatarsal flexion moment of the ankle to the ground. Previous studies have confirmed the contribution of passive ligament structure to this mechanism, but few studies have paid attention to the role of contractile components of longitudinal

arch in this process [129]. Through this research we found that with the increase of PF stiffness, the length of longitudinal arch gradually decreased, the height of longitudinal arch increased, the stiffness of PF increased from 25% to 200%, the length of longitudinal arch decreased by 4.14%, and the height of longitudinal arch increased by 7.02%. The windlass mechanism is caused by the dorsiflexion of MTPJ, especially the role of the first MTPJ [130]. The toes are dorsiflexed, and the PA is stretched around the metatarsal heads. The MLA decreased, and the plantar flexion angle between the calcaneus increased, which led to the increase of the foot arch [57]. This passive mechanism tightens the PF, thus stabilizing the longitudinal arch of the foot [52, 53], resulting in a rigid foot structure for pushing away [131].

In the previous biomechanical study of PF, the method of implanting sensors in vivo was often used. In order to examine the biomechanical response of PF under tension, Erdemir et al. [132] inserted sensors into the plantar aponeurosis and used body models to directly monitor the tension of PF during simulated walking. Carlson et al. [133] used sensors to measure the tension of the PF of the body and found that the PF force increased with the increase of toe dorsiflexion. However, these techniques can only measure the strain in a specific area, but not the distribution and direction of stress. The windlass mechanism mainly acts on the PF under the first metatarsal [134, 135], and the first metatarsal acts as a fulcrum in the propulsion process [136, 137]. The tension of the plantar aponeurosis leads to bone compression in the foot, effectively elevates the MLA and creates the stability of the mid-foot joint required for load-bearing forefoot propulsion [55, 56]. There are few studies on the mechanism of the foot transverse arch when the foot windlass mechanism occurs. This study found that with the increase of the PF stiffness, the foot longitudinal arch shortened, the height effectively increased, and the longitudinal arch stiffness was enhanced. But the stiffness of the transverse arch of the foot does not increase with the increase of the stiffness of the PF. The stress analysis of the PF shows that the flexion angle of the MTPJ decreases with the increase of the stiffness of the PF. As a result, the stress of the distal end of the PF is reduced, so the force of pulling the distal end of the metatarsal to the outside is reduced, and the proximal end of the

metatarsal is tightened to the medial side, resulting in a decrease in the stiffness of the posterior transverse arch of the foot. The FE model established in this study is a foot model with a flexion angle of 30 ° of the MTPJ, and the forefoot is in contact with the ground, when the flexion of the MTPJ is too small, the GRF further acts on the distal end of the metatarsals, and the distal ends of the metatarsals disperse outward. The proximal end tightens medially, resulting in an increase in the stiffness of the posterior transverse arch of the foot. Therefore, the PF stiffness affects the tension of the distal end of the PF by affecting the flexion of the MTPJ in the plantar windlass mechanism, and affects the stiffness of the transverse arch of the foot together with the GRF acting on the distal metatarsal. However, the stress at the middle and proximal end of PF increases with the increase of PF stiffness, which is consistent with the conclusion of Erdemir [132] and Carlson [133], but excessive and repeated load concentrated on PA is considered to be one of the most likely factors for the development of plantar fasciitis [138, 139].

4.2 Effect of Stair Height on Knee Biomechanics

This study developed an individualized, multimodal framework that integrates optimized biplanar fluoroscopy registration, subject-specific FE modeling, and musculoskeletal simulation to quantify knee joint stress during stair ascent. By combining DFIS-based in vivo kinematics, CT-derived anatomical models, and muscle forces estimated from OpenSim and validated against EMG, we established a quasi-dynamic FE model that reproduced reported cruciate ligament length changes and provided physiologically realistic stress distributions in the menisci and ligaments. The results showed that meniscal stresses increased systematically with stair height, with peak values concentrated in the body regions between 25% and 50% of the support phase and consistently higher magnitudes in the medial than the lateral meniscus. In addition, PCL stress peaked in the early support phase and increased markedly with stair height, whereas ACL stress peaked in mid-support and exhibited a more moderate height-dependent increase, indicating that taller stairs preferentially amplify early-stance PCL loading and mid-stance ACL loading.

This study showed that overall meniscal stress increased with stair height, especially during the transition from leg lift to foot contact (25%–50% of the gait cycle), where stress concentrations were greatest. This phase appears critical for meniscal loading due to rapid joint angle changes and heightened stability demands, leading to elevated central meniscal stress and emphasizing its role in shock absorption and load transmission [140]. At 200 mm, stress was more intense and widely distributed, with larger high-stress regions, reflecting the direct biomechanical impact of stair height [141]. Although medial and lateral meniscal stresses were similar before 25% stance, the medial meniscus subsequently carried consistently higher loads, suggesting greater vertical and shear loading over time [11], consistent with its thicker, less mobile anatomy and propensity for stress concentration [142]. Regionally, stress shifted from the posterior to the anterior horn, with maximum MB stress reaching 5.61 MPa at 200 mm, markedly exceeding other heights and indicating increased local injury risk. The MA region showed lower but gradually increasing stress that plateaued at 50%–75% of the cycle, likely reflecting full body-weight support during knee extension [143].

The lateral meniscus generally experienced lower stress than the medial meniscus, but its stress increased more rapidly with stair height. In particular, the LP region showed marked fluctuations between 25%–50% of the gait cycle, indicating sensitivity to stair height; although absolute stress was smaller, repetitive loading may still cause chronic strain [144]. Overall, stair height substantially affected meniscal loading, and high stairs (180–200 mm) produced both higher overall stress and more pronounced regional concentrations, suggesting an elevated risk of meniscal injury [145]. Thus, stair height should be carefully considered in real-world and rehabilitation settings, and high steps should be minimized for adults and individuals with knee pathology to reduce long-term meniscal damage.

Based on our FEA, increasing stair height systematically amplified peak stresses in both cruciate ligaments, but at different phases of the support period, which is consistent with current biomechanical understanding of stair climbing. At higher steps, greater knee flexion and larger extensor moments are required, especially in early stance, which

enhances posterior tibial shear and thereby increases PCL stress when the joint is more flexed; this agrees with musculoskeletal modeling showing that the PCL is one of the most heavily loaded ligaments during stair climbing and is particularly stressed under large flexion angles and strong hamstring activity[146]. In contrast, our finding that ACL stress peaks in mid-stance and rises with stair height is compatible with in vivo and modeling evidence that ACL loading is greatest when the knee approaches extension and the quadriceps PT complex produces anteriorly directed forces on the tibia [146]. A recent systematic review of in vivo ACL loading also reported a strong negative correlation between ACL elongation and knee flexion angle, with peak ACL strain typically occurring at smaller flexion angles and being further increased by greater anterior tibial shear and PT forces [147], which supports the pattern observed here at 50% of the support phase. Importantly, previous stair-climbing studies have highlighted stair negotiation as more mechanically demanding than level walking but have generally treated step geometry as a secondary factor; by explicitly varying stair height, our results extend this work and suggest that larger steps may disproportionately increase early-stance PCL loading while also elevating mid-stance ACL stress, which is relevant for setting safe stair dimensions and for prescribing stair-based rehabilitation or exercise in populations with cruciate ligament deficiency or reconstruction.

4.3 Effect of Jump Landing on the Knee Joint

This study integrates musculoskeletal simulation, FEA, and DFIS to examine the in vivo biomechanical state of the PT during the landing phase of stop-jump tasks. The key findings reveal that the maximum equivalent stress on the PT occurs at the initial moment of ground contact and decreases as knee flexion increases and the body decelerates. This finding is consistent with previous studies, which may explain why patients with patellar tendinopathy experience reduced tenderness and pain when flexing the knee beyond 60° [148]. However, activities which were highly related to patellar tendinopathy, such as volleyball and basketball, impose intense and unstable loads on the tendon. Therefore, the equivalent stress alone may not be sufficient to indicate the location of tendinopathy.

The maximum equivalent stress on the PT was consistently observed at the tendon's attachment sites with the tibia and patella. These regions are known to be prone to stress concentrations due to their role as transitional zones between soft tissue and bone, making them vulnerable to injury, particularly under repetitive high-load conditions as observed in athletes[149, 150]. Although clinical literature indicates that patellar tendinopathy most frequently occurs at the posterior proximal region surrounding the inferior pole of the patella[151, 152], the potential for injury at the distal insertion point should not be overlooked, particularly in high-impact sports. [153].

The strong correlation identified between knee flexion moments, quadriceps muscle forces, and maximum equivalent stress in the PT underscores the importance of these variables in tendon loading. The quadriceps play a crucial role in decelerating the body during landing, transmitting substantial forces through the PT, which results in elevated stress levels. This finding highlights the critical influence of knee flexion angle and velocity during landing on the loading environment of the PT, suggesting that excessive or uncontrolled knee flexion may increase the risk of tendinopathy or other overuse injuries[154]. Conversely, the observation that internal-external rotation and varus-valgus moments do not significantly impact maximum equivalent stress in the PT suggests that these rotational forces, while important for overall knee stability, are less influential in generating peak stresses within the tendon during landing. This implies that rotational stability, while vital for knee joint health, may not be the primary factor in preventing PT injuries during landing activities[155, 156].

The clinical implications of these findings are significant. Athletes, particularly those involved in high-impact sports such as basketball and volleyball, may benefit from targeted exercises designed to enhance the body's natural shock absorption capabilities. Plyometric training that emphasizes soft, controlled landings could mitigate the initial impact forces during landing, thereby reducing stress on the PT[157]. Additionally, the identification of stress concentrations at the PT's attachment sites suggests that these regions should be a focal point in injury prevention and rehabilitation programs.

However, it is essential to recognize its limitations. A major limitation is that the data was collected from only one participant, which may restrict the broader applicability of the results. The anatomical and biomechanical characteristics of the knee joint can vary significantly across individuals, expanding future research to include a broader and more varied group of participants would help validate and enhance these findings. Additionally, although the FEA model provides valuable insights into stress distribution within the PT, the simplifications of muscle forces and joint kinematics might influence the precision of the results. Future work should explore more complex models that incorporate these variables more accurately.

To improve understanding of PT stress variability across different groups, research should focus on developing models involving multiple participants. This approach will also aid in creating more tailored injury prevention and rehabilitation strategies. Moreover, integrating machine learning with biomechanical research presents significant opportunities. By analysing large datasets of biomechanical variables, machine learning techniques could help identify complex patterns and predict injury risk with greater accuracy. Integrating these techniques with FE models could lead to the development of real-time monitoring systems for athletes, providing timely feedback to prevent overuse injuries like tendinopathy.

4.4 Limation

The limitations of this study primarily concern sample representativeness and methodological assumptions. First, the model in each point is based on a single subject. While the single-subject design helps control confounding factors for within-subject comparisons, it cannot capture anatomical and biomechanical variability across populations. Therefore, the conclusions should be interpreted mainly as mechanistic insights and feasibility evidence, and broader generalization will require validation in larger and more diverse cohorts. Second, to maintain computational efficiency, the finite element model assumed isotropic, linear-elastic material behavior and simplified boundary conditions. These assumptions do not fully represent the nonlinear, anisotropic,

and viscoelastic characteristics of biological tissues, and may influence the absolute magnitudes and local distributions of the computed stresses, especially near insertion sites and contact regions. Thus, the future work should incorporate subject-specific material calibration and more physiologically detailed boundary conditions when feasible. Third, the quasi-static simulation framework overlooks time-dependent dynamic effects, such as inertia and muscle activation timing. In addition, muscle forces estimated from inverse dynamics and musculoskeletal modeling are subject to uncertainties arising from measurement noise and model assumptions, which can propagate into the finite element results.

5 CONCLUSIONS AND FUTURE WORKS

Fristly, DFIS can effectively explore the changes of foot arch during exercise and verify the effectiveness of the model. The effect of windlass mechanism increases with the enhancement of PF stiffness, and too low PF stiffness is not conducive to the stability of longitudinal arch. The stiffness of PF increases and the flexion angle of MTPJ decreases, which reduces the force of distal end of PF pulling the distal end of metatarsal bone to the outside, weakens the proximal end of metatarsal bone and weakens the medial tightening of metatarsal bone, which leads to the decrease of stiffness of transverse arch of foot; when the flexion angle of MTPJ is too small, the GRF acts on the distal end of metatarsal bone, which makes it disperse outward, tighten the proximal end inward, and increase the stiffness of transverse arch. Excessive stiffness of PF leads to stress concentration in the middle and proximal PF, which can easily lead to plantar fasciitis.

Secondly, this study proposed an individualized multimodal framework that integrates optimized biplanar fluoroscopy registration, subject-specific FE modeling, and musculoskeletal simulation to quantify knee joint stress during stair ascent. The quasi-dynamic FE model revealed that meniscal stress increased with stair height, with higher and more concentrated stresses in the medial meniscus, particularly in the body region around 25–50% of the support phase. PCL stress peaked in the early support phase and increased markedly with stair height, whereas ACL stress peaked in mid-support and showed a more moderate height-dependent increase, indicating that taller steps preferentially amplify early-stance PCL loading and mid-stance ACL loading. These findings underscore stair height as an important geometric factor influencing internal knee loading and provide biomechanical insight to support safer stair design and targeted rehabilitation strategies for adults.

Lastly the findings highlight that maximum equivalent stress in the PT is highest at the attachment sites during initial ground contact, decreasing as knee flexion increases throughout the landing phase. The strong correlation between knee flexion moments and quadriceps muscle forces with maximum equivalent stress emphasizes the critical role

these factors play in PT loading. Conversely, internal-external rotation and varus-valgus moments did not significantly impact PT stress. This suggests that rotational forces, while important for overall knee stability, are less influential in generating peak stresses within the tendon during landing. These insights could inform targeted interventions aimed at reducing the risk of patellar tendinopathy in athletes. However, it is important to note the study's limitation of using data from a single participant, which may restrict the generalizability of the findings. Future research should involve a larger and more diverse cohort to validate and extend these results. Additionally, integrating machine learning techniques with FE models could enhance the ability to predict injury risk and develop real-time monitoring systems for athletes, ultimately contributing to more personalized injury prevention and rehabilitation strategies.

NEW SCIENTIFIC THESIS POINTS

1st Thesis point

I combined measured data from a Dual Fluoroscopic Imaging System motion capture, and musculoskeletal modeling techniques to construct and validate a high-precision foot finite element model. By adjusting the stiffness of the plantar fascia (PF) (from 25% to 200% of the reference value), the study simulated and analyzed its mechanical effects on the plantar windlass mechanism and foot arch morphology (Figure 36).

- *Increasing plantar fascia stiffness does not linearly enhance the transverse arch. Transverse arch stiffness increases significantly only when PF stiffness is extremely high (>150%); whereas in the low-to-medium stiffness range, the transverse arch widens and exhibits reduced stiffness.*
- *The higher the plantar fascia stiffness, the smaller the dorsiflexion angle of the first metatarsophalangeal joint (MTPJ). This weakens the fascia's outward pull on the distal metatarsals, thereby altering the loading pattern and morphology of the forefoot.*
- *As stiffness increases, the stress distribution in the plantar fascia shifts: distal stress decreases, while stress in the middle and proximal sections rises significantly. This excessive stress concentration may be a key factor leading to plantar fasciitis.*

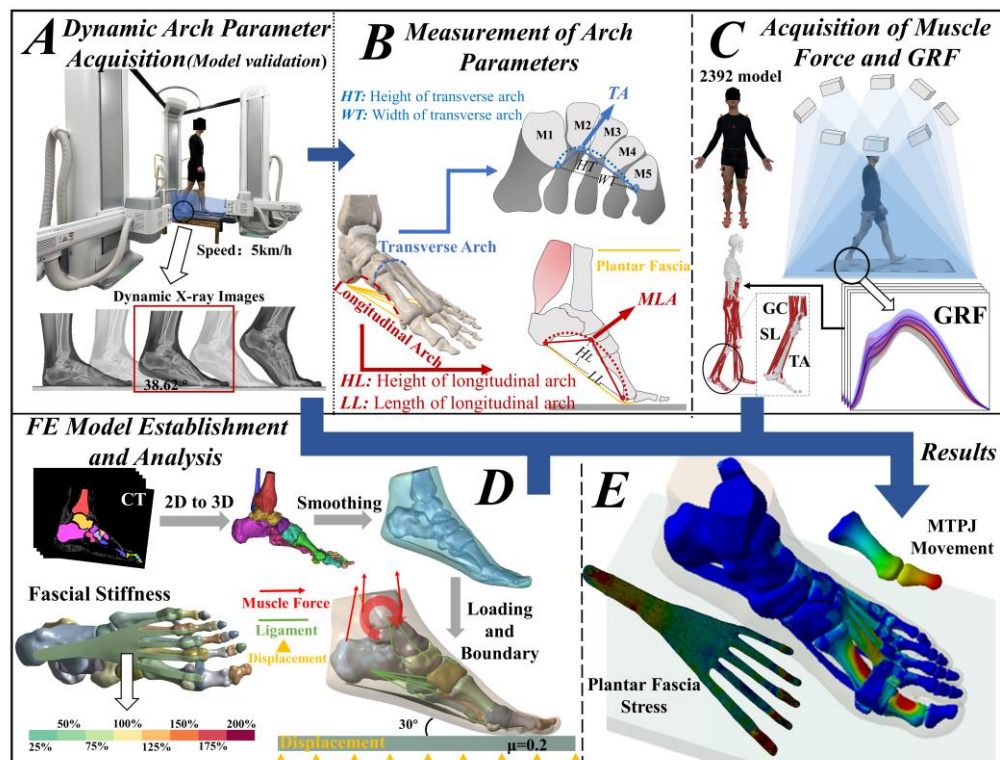


Figure 36. The flow of FE analysis and model verification. A: Uses DFIS to obtain dynamic foot data at 5km/h speed; B: Measures the parameters of longitudinal arch and transverse arch of foot; C: Obtains kinematic and dynamic data by Vicon system; D: Establishes FE model and uses muscle force data as load for FE analysis; E: Predicts foot arch data and stress distribution under different stiffness of PF.

● **Related articles to the first thesis point:**

Zhang, Q., Sun, D., Wang, M., Tafferner-Gulyás, V., Chen, H., Bíró, I., & Gu, Y. (2024). Effect of plantar fascia stiffness on plantar windlass mechanism and arch: Finite element method and dual fluoroscopic imaging system verification. *Medical Engineering & Physics*, 134, 104259.

Zhang, Q., Sun, D., Song, Y., Chen, H., Cen, X., Bíró, I., & Gu, Y. (2025). Effects of Plantar Fascia Stiffness on Windlass Mechanism: A Finite Elment Analysis. *Journal of medical biomechanics*, 40(1), 93-99.

2nd Thesis point

This study represents the first attempt to capture in vivo knee skeletal kinematics during stair ascent using a Dual Fluoroscopic Imaging System (DFIS). By mapping kinematic data onto a finite element model via image-model registration, a quasi-dynamic FE analysis was achieved. The workflow involved: acquiring high-precision skeletal trajectories via DFIS; constructing sophisticated 3D anatomical models based on CT imagery; and estimating muscle forces through OpenSim inverse dynamics using Vicon motion capture and force plate data, validated by electromyography. These parameters were ultimately utilized to drive a tibiofemoral joint FE model, enabling the quantitative evaluation of stress distributions in the menisci, anterior cruciate ligaments, and posterior cruciate ligaments at critical time points of the stance phase (Figure 37).

- *The study found a systematic upward trend in overall meniscal stress with increasing stair height. The most distinct stress concentration occurred in the body region of the medial meniscus, particularly during the 25% to 50% phase of the stance phase (the transition from leg lift to full foot contact). At the maximum stair height of 200 mm, the peak stress in the medial meniscus body was significantly higher than in other regions and at other heights.*
- *The stress in the posterior cruciate ligament (PCL) reached its primary peak in the early stance phase (approximately at the 25% time point). Furthermore, the magnitude of PCL stress increase showed a significant positive correlation with stair height (increasing from 0.65 MPa at 120 mm to 1.32 MPa at 200 mm), indicating that higher stairs preferentially and substantially amplify PCL loading during early stance.*
- *Unlike the PCL, stress in the anterior cruciate ligament (ACL) peaked during the mid-stance phase (approximately at the 50% time point). Although peak ACL stress also increased with stair height, the rate of growth was more moderate compared to the PCL, suggesting that the ACL primarily bears load during mid-stance and is slightly less sensitive to changes in stair height than the PCL.*

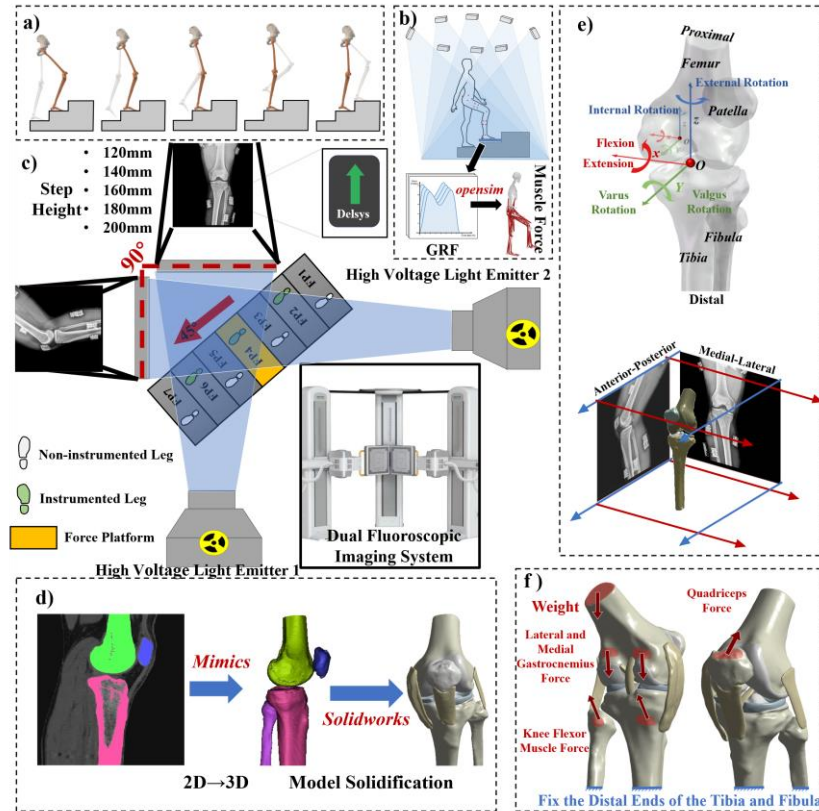


Figure 37. Workflow of the quasi-dynamic finite element model establishment and analysis. a) Single-leg support phase. b) Kinematic and dynamic data obtained through the Vicon 3D motion capture system and portable 3D force plate to calculate muscle force. c) Dynamic X-ray images captured during stair climbing using the DFIS. d) 3D model established based on the subject's CT imaging data. e) Registration of 3D Models with Dynamic Images and Establishment of Coordinate Systems. f) Finite element boundaries and loads.

● **Related articles to the first thesis point:**

Zhang, Q., Xu, Y., Sun, D., Chen, H., Sárosi, J., Bíró I., & GU, Y. Individualized Multimodal Integration of Optimized Dual Fluoroscopic Imaging Registration and Finite Element Modeling for Knee Joint Stress During Stair Ascent. *Medical & Biological Engineering & Computing*, under review.

Zhang, Q., Chen, H., Bíró, I., & Fekete, G. (2024). Biomechanical Characteristics of Adolescent Cervical Forward Flexion Analyses Based on the Finite Element Method. In *Material Strength and Applied Mechanics* (pp. 479-485). IOS Press.

3rd Thesis point

In this study, high-speed dynamic X-ray images of a healthy male subject during the landing phase of a stop-jump task were acquired using a Dual Fluoroscopic Imaging System (DFIS). By combining data from a Vicon motion capture system, force plates, and electromyography (EMG), musculoskeletal modeling was performed via OpenSim to calculate joint moments and muscle forces. These data were used to drive an individualized knee joint finite element (FE) model, which was precisely calibrated using DFIS, to conduct transient stress analysis at six key moments during the landing process. Additionally, ridge regression analysis was employed to explore the relationship between biomechanical loads and the maximum equivalent (von Mises) stress in the patellar tendon (PT) (Figure 38).

- *Finite element analysis revealed that throughout the landing phase, the maximum equivalent stress in the patellar tendon consistently occurred at its attachment sites to the tibia or patella. Most critically, the peak stress (reaching 94.44 MPa) appeared at the initial instant of ground contact, after which the stress gradually decreased to a minimum (16.37 MPa) as the knee flexion angle increased and the body decelerated.*
- *Ridge regression analysis demonstrated a significant positive correlation between the knee flexion moment and quadriceps muscle force with the maximum equivalent stress in the patellar tendon (standardized coefficients of 0.747 and 0.181, respectively; $P < 0.001$). This indicates that these two factors are the key drivers determining patellar tendon loading during landing.*
- *The study found that although internal/external rotation moments and varus/valgus moments are important for overall joint stability, they had no significant effect on the maximum equivalent stress of the patellar tendon. This implies that for the prevention of patellar tendon injury during landing, controlling rotational forces may not be the primary consideration; instead, attention should be focused more on load management in the flexion plane.*

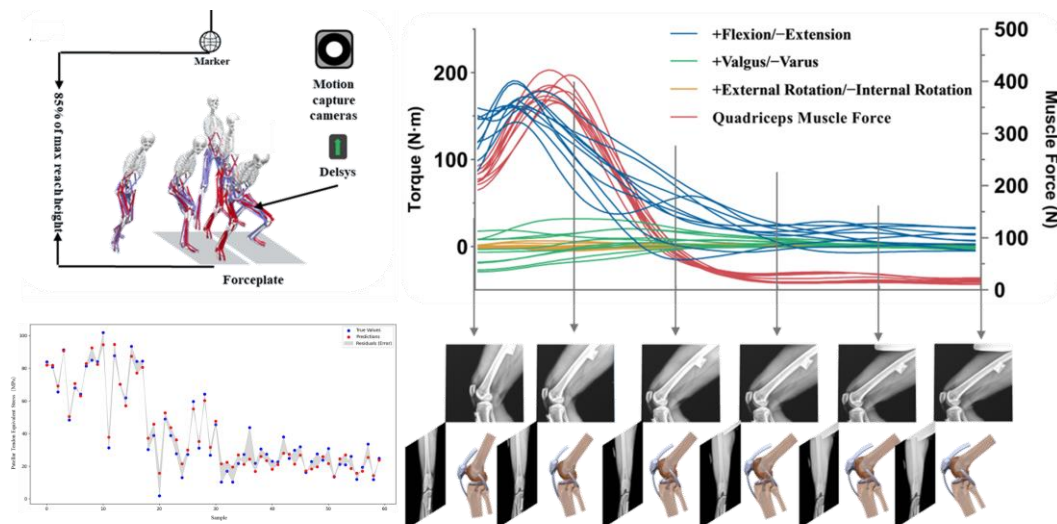


Figure 38. Diagram of data acquisition, modeling, and statistical analysis. Data collection of kinematics and kinetics during stop-jump tasks; Application of kinetic data to the aligned model. The figure also presents the Ridge regression analysis of three-dimensional knee joint moments and quadriceps muscle forces influencing maximum equivalent stress in the patellar tendon during the landing phase.

- **Related articles to the first thesis point:**

Li, F., Sun, D., Song, Y., Fang, Y., Cen, X., **Zhang, Q.**, & Gu, Y. (2025). Comparison of Landing Biomechanics in Male Amateur Basketball Players with and without Patellar Tendinopathy during Simulated Games. *Journal of Human Kinetics*, 96(Spec Issue), 69.

Li, F., Sun, D., Song, Y., Zhou, Z., Wang, D., Cen, X., **Zhang, Q.**, Gao Z., & Gu, Y. (2025). Dynamic simulation of knee joint mechanics: individualized multi-moment finite element modelling of patellar tendon stress during landing. *Journal of Biomechanics*, 112730.

LIST OF PUBLICATIONS

1. **Zhang, Q.**, Zhang, Y., Chon, T. E., Baker, J. S., & Gu, Y. (2023). Analysis of stress and stabilization in adolescent with osteoporotic idiopathic scoliosis: finite element method. *Computer Methods in Biomechanics and Biomedical Engineering*, 26(1), 12-24.
2. **Zhang, Q.**, Sun, D., Wang, M., Tafferner-Gulyás, V., Chen, H., Bíró, I., & Gu, Y. (2024). Effect of plantar fascia stiffness on plantar windlass mechanism and arch: Finite element method and dual fluoroscopic imaging system verification. *Medical Engineering & Physics*, 134, 104259.
3. **Zhang, Q.**, Sun, D., Song, Y., Chen, H., Cen, X., Bíró, I., & Gu, Y. (2025). Effects of Plantar Fascia Stiffness on Windlass Mechanism: A Finite Element Analysis. *Journal of medical biomechanics*, 40(1), 93-99.
4. **Zhang, Q.**, & Bíró, I. (2024). Biomechanical analysis and modeling of different traction patterns in adolescent idiopathic scoliosis. *Analecta Technica Szegedinensia*, 18(3), 88-100.
5. **Zhang, Q.**, Chen, H., & Bíró, I. (2024, October). Effect of stiffness difference of forefoot transverse arch on foot biomechanics. In *2024 IEEE 7th International Conference and Workshop Óbuda on Electrical and Power Engineering (CANDO-EPE)* (pp. 53-58). IEEE.
6. **Zhang, Q.**, Chen, H., Bíró, I., & Fekete, G. (2024). Biomechanical Characteristics of Adolescent Cervical Forward Flexion Analyses Based on the Finite Element Method. In *Material Strength and Applied Mechanics* (pp. 479-485). IOS Press.
7. **Zhang, Q.**, Chen, H., Jiang, X., & Bíró, I. (2024, May). Prediction and Effectiveness Verification of Chevron Osteotomy Surgical Plan Using Finite Element Method. In *2024 IEEE 18th International Symposium on Applied Computational Intelligence and Informatics (SACI)* (pp. 000089-000094). IEEE.

8. **Zhang, Q.**, & Bíró, I. (2024). Effects of different schemes of hallux valgus osteotomy on the stress based on computer digital simulation. In R. Horváth, J. Lukács, & R. G. Stadler (Eds.), *Mérnöki Szimpózium a Bánkin: Proceedings of the Engineering Symposium at Bánki (ESB 2024)* (pp. 97–102). Obuda University.
9. Zhang, L., **Zhang, Q.**, Liu, Q., Jiang, X., Simon, J., Hortobágyi, T., & Gu, Y. (2025). Dual Transverse Arch Foot Orthosis Improves Gait Biomechanics in Females with Flexible Flatfoot. *Bioengineering*, 12(4), 418.
10. Zhang, L., **Zhang, Q.**, Zhong, Y., Hortobágyi, T., & Gu, Y. (2024). Effect of forefoot transverse arch stiffness on foot biomechanical response--based on finite element method. *Frontiers in Bioengineering and Biotechnology*, 12, 1387768.
11. Li, F., Sun, D., **Zhang, Q.**, Chen, H., Bíró, I., Zheng, Z., & Gu, Y. (2023). Effect of a Double Helical Spring Decompression Structure Backpack on the Lumbar Spine Biomechanics of School-Age Children: A Finite Element Study. *Molecular & Cellular Biomechanics*, 20(1).
12. Jiang, X., **Zhang, Q.**, & Bíró, I. (2024, May). Effects of Long-Distance Running on Lower Limb Joint Kinematics in Recreational Runners. In 2024 IEEE 18th International Symposium on Applied Computational Intelligence and Informatics (SACI) (pp. 000095-000098). IEEE.
13. Zhang, C., Song, Y., **Zhang, Q.**, Teo, E. C., & Liu, W. (2024). Biomechanical study of symmetric bending and lifting behavior in weightlifter with lumbar L4-L5 disc herniation and physiological straightening using finite element simulation. *Bioengineering*, 11(8), 825.
14. Lu, Z., Li, X., Gao, Z., Chen, H., **Zhang, Q.**, Xu, Y., ... & Gu, Y. (2024). Customized 3D-Printed Insoles for Diabetic Foot Care: Finite Element Analysis and Machine Learning Approach. In *Material Strength and Applied Mechanics* (pp. 515-522). IOS Press.
15. Li, F., Sun, D., Song, Y., Fang, Y., Cen, X., **Zhang, Q.**, & Gu, Y. (2025). Comparison

- of Landing Biomechanics in Male Amateur Basketball Players with and without Patellar Tendinopathy during Simulated Games. *Journal of Human Kinetics*, 96(Spec Issue), 69.
16. Chen, H., Sun, D., **Zhang, Q.**, Liu, Q., Bíró, I., & Gu, Y. (2024). Biomechanical Differences Between Habitual Forefoot and Rearfoot Strike Running: A Systematic Review. *Advanced Exercise and Health Science*.
 17. Chen, H., **Zhang, Q.**, & Bíró, I. (2024, October). Effects of Arch Support Pads and Insoles on Gait Parameters and Plantar Mechanics During Running in Adults with Flatfoot. In *2024 IEEE 7th International Conference and Workshop Óbuda on Electrical and Power Engineering (CANDO-EPE)* (pp. 000099-000102). IEEE.
 18. Chen, H., **Zhang, Q.**, Bíró, I., & Fekete, G. (2024). Effects of Different Longitudinal Bending Stiffness Shoes on Distal Joint Kinematics and Muscular Mechanics in Adolescent Amateur Runners. In *Material Strength and Applied Mechanics* (pp. 472-478). IOS Press.
 19. Chen, H., Sun, D., Fang, Y., Gao, S., **Zhang, Q.**, Bíró, I., ... & Gu, Y. (2024). Effect of orthopedic insoles on lower limb motion kinematics and kinetics in adults with flat foot: a systematic review. *Frontiers in Bioengineering and Biotechnology*, 12, 1435554.
 20. Li, F., Sun, D., Song, Y., Zhou, Z., Wang, D., Cen, X., **Zhang, Q.**, Gao Z., & Gu, Y. (2025). Dynamic simulation of knee joint mechanics: individualized multi-moment finite element modelling of patellar tendon stress during landing. *Journal of Biomechanics*, 112730.
 21. Gao, S., Sun, D., Song, Y., Cen, X., **Zhang, Q.**, Gao, Z., ... & Gu, Y. (2025). Effect of foot-shaped bionic shoes on ground reaction forces and foot stress at various running speeds. *Computer Methods in Biomechanics and Biomedical Engineering*, 1-11.
 22. Li, F., Sun, D., Zhu, C., **Zhang, Q.**, Song, Y., Cen, X., ... & Gu, Y. (2024). The impact

- of toe spring and foot strike angle on forefoot running biomechanics: a finite element analysis. *Computer Methods in Biomechanics and Biomedical Engineering*, 1-11.
23. Zhu, A., Gao, S., Huang, L., Chen, H., **Zhang, Q.**, Sun, D., & Gu, Y. (2024). Effects of fatigue and unanticipated factors on knee joint biomechanics in female basketball players during cutting. *Sensors (Basel, Switzerland)*, 24(14), 4759.
24. Chen, H., **Zhang, Q.**, & Biró, I. (2024, May). A Comparative Analysis of Different Longitudinal Bending Stiffness Shoes on Running Biomechanics in Adolescent Amateur Runners. In *2024 IEEE 18th International Symposium on Applied Computational Intelligence and Informatics (SACI)* (pp. 000083-000088). IEEE.
25. Cai, J., Sun, D., Xu, Y., Chen, H., **Zhang, Q.**, Baker, J. S., & Gu, Y. (2024). The influence of medial and lateral forefoot height discrepancy on lower limb biomechanical characteristics during the stance phase of running. *Applied Sciences*, 14(13), 5807.

REFERENCES

- [1] S. Han, S.J. Son, H. Kim, H. Lee, M. Seeley, T. Hopkins, Prelanding movement strategies among chronic ankle instability, coper, and control subjects, *Sports Biomechanics*, 21 (2022) 391-407.
- [2] S. Kuratani, Evolution of the vertebrate jaw: comparative embryology and molecular developmental biology reveal the factors behind evolutionary novelty, *Journal of anatomy*, 205 (2004) 335-347.
- [3] T. Ohnishi, M. Suzuki, A. Nawata, S. Naomoto, T. Iwasaki, H. Haneishi, Three-dimensional motion study of femur, tibia, and patella at the knee joint from bi-plane fluoroscopy and CT images, *Radiological physics and technology*, 3 (2010) 151-158.
- [4] R. De Ridder, T. Willems, J. Vanrenterghem, M.A. Robinson, T. Palmans, P. Roosen, Multi-segment foot landing kinematics in subjects with chronic ankle instability, *Clinical Biomechanics*, 30 (2015) 585-592.
- [5] C.C. Herb, K. Grossman, M.A. Feger, L. Donovan, J. Hertel, Lower extremity biomechanics during a drop-vertical jump in participants with or without chronic ankle instability, *Journal of athletic training*, 53 (2018) 364-371.
- [6] D.A. Bruening, K.M. Cooney, F.L. Buczek, Analysis of a kinetic multi-segment foot model. Part I: Model repeatability and kinematic validity, *Gait & posture*, 35 (2012) 529-534.
- [7] R.M. Koldenhoven, M.A. Feger, J.J. Fraser, S. Saliba, J. Hertel, Surface electromyography and plantar pressure during walking in young adults with chronic ankle instability, *Knee Surgery, Sports Traumatology, Arthroscopy*, 24 (2016) 1060-1070.
- [8] E.A. Wikstrom, C.N. Brown, Minimum reporting standards for copers in chronic ankle instability research, *Sports Medicine*, 44 (2014) 251-268.
- [9] A. Planta, R. Surbeck, W.R. Taylor, S. Plüss, F. Vogl, P. Schütz, A dual-plane fluoroscope to track joint kinematics during dynamic daily activities, *PloS one*, 20 (2025)

e0328351.

- [10] L.A. Kelly, A.G. Cresswell, S. Racinais, R. Whiteley, G. Lichtwark, Intrinsic foot muscles have the capacity to control deformation of the longitudinal arch, *Journal of The Royal Society Interface*, 11 (2014) 20131188.
- [11] S. Park, S. Lee, J. Yoon, S.W. Chae, Finite element analysis of knee and ankle joint during gait based on motion analysis, *Med Eng Phys*, 63 (2019) 33-41.
- [12] P. Lundgren, C. Nester, A. Liu, A. Arndt, R. Jones, A. Stacoff, P. Wolf, A. Lundberg, Invasive in vivo measurement of rear-, mid-and forefoot motion during walking, *Gait & posture*, 28 (2008) 93-100.
- [13] F.E. Zajac, Muscle coordination of movement: a perspective, *Journal of biomechanics*, 26 (1993) 109-124.
- [14] S.J. Hall, *Basic biomechanics*, McGraw-Hill Higher Education 2021.
- [15] D. Knudson, *Fundamentals of Biomechanics and Qualitative Movement Diagnosis*, *Fundamentals of Biomechanics*, Springer 2021, pp. 37-51.
- [16] P.X. Fuchs, A. Fusco, J.W. Bell, S.P. von Duvillard, C. Cortis, H. Wagner, Effect of Differential Training on Female Volleyball Spike-Jump Technique and Performance, *Int J Sports Physiol Perform*, (2020) 1-7.
- [17] M. Carson, M. Harrington, N. Thompson, J. O'connor, T. Theologis, Kinematic analysis of a multi-segment foot model for research and clinical applications: a repeatability analysis, *Journal of biomechanics*, 34 (2001) 1299-1307.
- [18] Y. Sekiguchi, T. Kokubun, H. Hanawa, H. Shono, A. Tsuruta, N. Kanemura, Evaluation of the validity, reliability, and kinematic characteristics of multi-segment foot models in motion capture, *Sensors*, 20 (2020) 4415.
- [19] N.A. Sharkey, A.J. Hamel, A dynamic cadaver model of the stance phase of gait: performance characteristics and kinetic validation, *Clinical Biomechanics*, 13 (1998) 420-433.

- [20] A. Arndt, P. Westblad, I. Winson, T. Hashimoto, A. Lundberg, Ankle and subtalar kinematics measured with intracortical pins during the stance phase of walking, *Foot & ankle international*, 25 (2004) 357-364.
- [21] S. Guan, H.A. Gray, F. Keynejad, M.G. Pandy, Mobile biplane X-ray imaging system for measuring 3D dynamic joint motion during overground gait, *IEEE transactions on medical imaging*, 35 (2015) 326-336.
- [22] F. Li, D. Sun, Y. Song, Z. Zhou, D. Wang, X. Cen, Q. Zhang, Z. Gao, Y. Gu, Dynamic simulation of knee joint mechanics: individualized multi-moment finite element modelling of patellar tendon stress during landing, *Journal of Biomechanics*, (2025) 112730.
- [23] H.A. Gray, S. Guan, L.T. Thomeer, A.G. Schache, R. de Steiger, M.G. Pandy, Three-dimensional motion of the knee-joint complex during normal walking revealed by mobile biplane x-ray imaging, *Journal of Orthopaedic Research®*, 37 (2019) 615-630.
- [24] J.A. Cross, B.D. McHenry, R. Molthen, E. Exten, T.G. Schmidt, G.F. Harris, Biplane fluoroscopy for hindfoot motion analysis during gait: A model-based evaluation, *Medical Engineering & Physics*, 43 (2017) 118-123.
- [25] M.J. Bey, R. Zael, S.K. Brock, S. Tashman, Validation of a new model-based tracking technique for measuring three-dimensional, in vivo glenohumeral joint kinematics, *Journal of biomechanical engineering*, 128 (2006) 604-609.
- [26] L.B. Chou, R.A. Mann, M.M. Casillas, Biplanar chevron osteotomy, *Foot & ankle international*, 19 (1998) 579-584.
- [27] C.D. Untaroiu, N. Yue, J. Shin, A finite element model of the lower limb for simulating automotive impacts, *Annals of biomedical engineering*, 41 (2013) 513-526.
- [28] L. Grassi, S.P. Väänänen, L. Jelpsson, Ö. Ljunggren, B.E. Rosengren, M.K. Karlsson, H. Isaksson, 3D Finite Element Models Reconstructed From 2D Dual-Energy X-Ray

Absorptiometry (DXA) Images Improve Hip Fracture Prediction Compared to Areal BMD in Osteoporotic Fractures in Men (MrOS) Sweden Cohort, *Journal of Bone and Mineral Research*, 38 (2023) 1258-1267.

[29] R. Awal, M. Naznin, T.R. Faisal, Machine learning based finite element analysis (FEA) surrogate for hip fracture risk assessment and visualization, *Expert Systems with Applications*, 264 (2025) 125916.

[30] R. Verma, K.K. Pandey, S. Tripathi, H. Ahirwar, H.S. Nanda, 3D Modeling of CF-PEEK Bone Plates for Tibia Fixation, *Novel Applications of Functionally Graded Materials*, CRC Press, pp. 202-218.

[31] Y. Xiao, M. Ling, Z. Liang, J. Ding, S. Zhan, H. Hu, B. Chen, Dual fluoroscopic imaging and CT-based finite element modelling to estimate forces and stresses of grafts in anatomical single-bundle ACL reconstruction with different femoral tunnels, *International Journal of Computer Assisted Radiology and Surgery*, 16 (2021) 495-504.

[32] A. Asghar, S. Naaz, The transverse arch in the human feet: A narrative review of its evolution, anatomy, biomechanics and clinical implications, *Morphologie*, 106 (2022) 225-234.

[33] J.R. Jastifer, L.H. Toledo-Pereyra, Leonardo da Vinci's Foot: historical evidence of concept, *Journal of Investigative Surgery*, 25 (2012) 281-285.

[34] N.B. Holowka, D.E. Lieberman, Rethinking the evolution of the human foot: insights from experimental research, *Journal of experimental biology*, 221 (2018) jeb174425.

[35] D.A. Neumann, E.R. Kelly, *Kinesiology of the musculoskeletal system: foundations for rehabilitation*, (2010).

[36] N.B. Holowka, A. Richards, B.E. Sibson, D.E. Lieberman, The human foot functions like a spring of adjustable stiffness during running, *Journal of Experimental Biology*, 224 (2021) jeb219667.

[37] S. Raychoudhury, D. Hu, L. Ren, Three-dimensional kinematics of the human

metatarsophalangeal joint during level walking, *Frontiers in bioengineering and biotechnology*, 2 (2014) 73.

[38] A. Boonpratotong, L. Ren, The human ankle-foot complex as a multi-configurable mechanism during the stance phase of walking, *Journal of Bionic Engineering*, 7 (2010) 211-218.

[39] H. Gong, C. Zhang, Y. Sui, Contact-Rich and Deformable Foot Modeling for Locomotion Control of the Human Musculoskeletal System, 2025 IEEE-RAS 24th International Conference on Humanoid Robots (Humanoids), IEEE, 2025, pp. 301-308.

[40] C. Nester, R.K. Jones, A. Liu, D. Howard, A. Lundberg, A. Arndt, P. Lundgren, A. Stacoff, P. Wolf, Foot kinematics during walking measured using bone and surface mounted markers, *Journal of biomechanics*, 40 (2007) 3412-3423.

[41] C. Maiwald, A. Arndt, C. Nester, R. Jones, A. Lundberg, P. Wolf, The effect of intracortical bone pin application on kinetics and tibiocalcaneal kinematics of walking gait, *Gait & Posture*, 52 (2017) 129-134.

[42] C.-B. Phan, G. Shin, K.M. Lee, S. Koo, Skeletal kinematics of the midtarsal joint during walking: Midtarsal joint locking revisited, *Journal of Biomechanics*, 95 (2019) 109287.

[43] M. Balsdon, C. Dombroski, K. Bushey, T.R. Jenkyn, Hard, soft and off-the-shelf foot orthoses and their effect on the angle of the medial longitudinal arch: A biplane fluoroscopy study, *Prosthetics and orthotics international*, 43 (2019) 331-338.

[44] C.-B. Phan, K.M. Lee, S.-S. Kwon, S. Koo, Kinematic instability in the joints of flatfoot subjects during walking: A biplanar fluoroscopic study, *Journal of Biomechanics*, 127 (2021) 110681.

[45] F.G. Evans, S. Band, *Studies on the Anatomy and Function of Bone and Joints*, Springer 1966.

[46] M. Nordin, V.H. Frankel, *Basic biomechanics of the musculoskeletal system*,

Lippincott Williams & Wilkins 2001.

[47] T.B. Belytschko, T.P. Andriacchi, A.B. Schultz, J.O. Galante, Analog studies of forces in the human spine: computational techniques, *J Biomech*, 6 (1973) 361-371.

[48] R.E. Tannous, F.A. Bandak, T.G. Toridis, R.H. Eppinger, A three-dimensional finite element model of the human ankle: development and preliminary application to axial impulsive loading, *SAE Technical Paper*, 1996.

[49] E. Morales-Orcajo, T.R. Souza, J. Bayod, E. Barbosa de Las Casas, Non-linear finite element model to assess the effect of tendon forces on the foot-ankle complex, *Med Eng Phys*, 49 (2017) 71-78.

[50] Y. Zhang, J. Awrejcewicz, O. Szymanowska, S. Shen, X. Zhao, J.S. Baker, Y. Gu, Effects of severe hallux valgus on metatarsal stress and the metatarsophalangeal loading during balanced standing: A finite element analysis, *Computers in biology and medicine*, 97 (2018) 1-7.

[51] N.B. Holowka, M.C. O'Neill, N.E. Thompson, B. Demes, Chimpanzee and human midfoot motion during bipedal walking and the evolution of the longitudinal arch of the foot, *Journal of Human Evolution*, 104 (2017) 23-31.

[52] R. Ker, M. Bennett, S. Bibby, R. Kester, R.M. Alexander, The spring in the arch of the human foot, *Nature*, 325 (1987) 147-149.

[53] A.N. Natali, P.G. Pavan, C. Stecco, A constitutive model for the mechanical characterization of the plantar fascia, *Connective Tissue Research*, 51 (2010) 337-346.

[54] G.A. Arangio, C. Chen, E.P. Salathé, Effect of varying arch height with and without the plantar fascia on the mechanical properties of the foot, *Foot & ankle international*, 19 (1998) 705-709.

[55] E.R. Hageman, M. Hall, E.G. Sterner, G.A. Mirka, Medial longitudinal arch deformation during walking and stair navigation while carrying loads, *Foot & ankle international*, 32 (2011) 623-629.

- [56] N.L. Griffin, C.E. Miller, D. Schmitt, K. D'Août, Understanding the evolution of the windlass mechanism of the human foot from comparative anatomy: Insights, obstacles, and future directions, *American journal of physical anthropology*, 156 (2015) 1-10.
- [57] J.H. Hicks, The mechanics of the foot. II. The plantar aponeurosis and the arch, *Journal of anatomy*, 88 (1954) 25-30.
- [58] H.-Y.K. Cheng, C.-L. Lin, S.-W. Chou, H.-W. Wang, Nonlinear finite element analysis of the plantar fascia due to the windlass mechanism, *Foot & ankle international*, 29 (2008) 845-851.
- [59] M. Venkadesan, A. Yawar, C.M. Eng, M.A. Dias, D.K. Singh, S.M. Tommasini, A.H. Haims, M.M. Bandi, S. Mandre, Stiffness of the human foot and evolution of the transverse arch, *Nature*, 579 (2020) 97-100.
- [60] D. Babu, B. Bordoni, *Anatomy, bony pelvis and lower limb, medial longitudinal arch of the foot*, (2020).
- [61] S. Li, Y. Zhang, Y. Gu, J. Ren, Stress distribution of metatarsals during forefoot strike versus rearfoot strike: A finite element study, *Computers in biology and medicine*, 91 (2017) 38-46.
- [62] H. Zhou, D. Xu, W. Quan, U.C. Ugbolue, Y. Gu, Effects of different contact angles during forefoot running on the stresses of the foot bones: a finite element simulation study, *Frontiers in bioengineering and biotechnology*, 12 (2024) 1337540.
- [63] D. Ye, X. Sun, C. Zhang, S. Zhang, X. Zhang, W. Fu, [Progress on in vivo ankle biomechanics based on dual fluoroscopic imaging technology], *Sheng wu yi xue gong cheng xue za zhi = Journal of biomedical engineering = Shengwu yixue gongchengxue zazhi*, 38 (2021) 602-608.
- [64] K.E. Roach, B. Wang, A.L. Kapron, N.M. Fiorentino, C.L. Saltzman, K. Bo Foreman, A.E. Anderson, In vivo kinematics of the tibiotalar and subtalar joints in asymptomatic subjects: a high-speed dual fluoroscopy study, *Journal of biomechanical engineering*, 138 (2016) 091006.

- [65] V. Kefala, Assessment of Normal Knee Kinematics using High-Speed Stereo-Radiography System, University of Denver, 2015.
- [66] T. Zhou, Y. Xu, L. Zhou, S. Wang, S. Wang, W. Xu, Multi-planar instability, laxity and reduced knee flexion during the support phase of walking are determinants of return to sports, *Frontiers in bioengineering and biotechnology*, 10 (2022) 1047135.
- [67] J.J. Elias, A.J. Cosgarea, Computational Models for Investigating the Patellofemoral Joint: Clinical Relevance, *Anterior Knee Pain and Patellar Instability*, Springer 2011, pp. 287-297.
- [68] M.A. Baldwin, C.W. Clary, C.K. Fitzpatrick, J.S. Deacy, L.P. Maletsky, P.J. Rullkoetter, Dynamic finite element knee simulation for evaluation of knee replacement mechanics, *Journal of biomechanics*, 45 (2012) 474-483.
- [69] M. Yan, T. Liang, H. Zhao, Y. Bi, T. Wang, T. Yu, Y. Zhang, Model properties and clinical application in the finite element analysis of knee joint: a review, *Orthopaedic surgery*, 16 (2024) 289-302.
- [70] A. Paz, G.A. Orozco, R.K. Korhonen, J.J. Garcia, M.E. Mononen, Expediting finite element analyses for subject-specific studies of knee osteoarthritis: a literature review, *Applied Sciences*, 11 (2021) 11440.
- [71] S. Chokhandre, A. Schwartz, E. Klonowski, B. Landis, A. Erdemir, Open knee (s): a free and open source library of specimen-specific models and related digital assets for finite element analysis of the knee joint, *Annals of biomedical engineering*, 51 (2023) 10-23.
- [72] L. Dagneaux, F. Canovas, F. Jourdan, Finite element analysis in the optimization of posterior-stabilized total knee arthroplasty, *Orthopaedics & Traumatology: Surgery & Research*, 110 (2024) 103765.
- [73] A. Karimi Dastgerdi, A. Esrafilian, C.P. Carty, A. Nasser, A. Yahyaiee Babil, M. Barzan, R.K. Korhonen, I. Astori, W. Hall, D.J. Saxby, Validation and evaluation of subject-specific finite element models of the pediatric knee, *Scientific reports*, 13 (2023)

18328.

[74] S.P. Magnusson, M.V. Narici, C.N. Maganaris, M. Kjaer, Human tendon behaviour and adaptation, in vivo, *The Journal of physiology*, 586 (2008) 71-81.

[75] I.S. Albers, J. Zwerver, R.L. Diercks, J.H. Dekker, I. Van den Akker-Scheek, Incidence and prevalence of lower extremity tendinopathy in a Dutch general practice population: a cross sectional study, *BMC musculoskeletal disorders*, 17 (2016) 16.

[76] I. Janssen, J.R. Steele, B.J. Munro, N.A. Brown, Predicting the patellar tendon force generated when landing from a jump, *Medicine and science in sports and exercise*, 45 (2013) 927-934.

[77] D.P. Richards, S.V. Ajemian, J.P. Wiley, R.F. Zernicke, Knee joint dynamics predict patellar tendinitis in elite volleyball players, *The American journal of sports medicine*, 24 (1996) 676-683.

[78] H.-T. Lin, W.-C. Kuo, Y. Chen, T.-Y. Lo, Y.-I. Li, J.-H. Chang, Effects of Fatigue in Lower Back Muscles on Basketball Jump Shots and Landings, *Physical Activity and Health*, (2022).

[79] D.S. Williams, I.S. McClay, Measurements used to characterize the foot and the medial longitudinal arch: reliability and validity, *Physical therapy*, 80 (2000) 864-871.

[80] A.C. Redmond, J. Crosbie, R.A. Ouvrier, Development and validation of a novel rating system for scoring standing foot posture: the Foot Posture Index, *Clinical biomechanics*, 21 (2006) 89-98.

[81] Q. Mei, Y. Gu, Foot pronation contributes to altered lower extremity loading after long distance running, *Frontiers in physiology*, 10 (2019) 454088.

[82] F. Mo, Y. Li, J. Li, S. Zhou, Z. Yang, A three-dimensional finite element foot-ankle model and its personalisation methods analysis, *International Journal of Mechanical Sciences*, 219 (2022) 107108.

[83] L.T. Atkins, M. Lowrey, S. Reagor, K. Walker, D. Cage, The effects of increasing

trunk flexion during stair ascent on the rate and magnitude of achilles tendon force in asymptomatic females, *Journal of Applied Biomechanics*, 39 (2022) 10-14.

[84] S.J. Warden, C.S. Wright, R.K. Fuchs, Bone Microarchitecture and Strength Adaptation to Physical Activity: A Within-Subject Controlled HRpQCT Study, *Medicine and science in sports and exercise*, 53 (2021) 1179-1187.

[85] H. Zhou, D. Xu, W. Quan, U.C. Ugbole, Z. Zhou, Y. Gu, Can the Entire Function of the Foot Be Concentrated in the Forefoot Area during the Running Stance Phase? A Finite Element Study of Different Shoe Soles, *Journal of Human Kinetics*, 92 (2024).

[86] M. Wang, S. Li, E.-C. Teo, G. Fekete, Y. Gu, The influence of heel height on strain variation of plantar fascia during high heel shoes walking-combined musculoskeletal modeling and finite element analysis, *Frontiers in bioengineering and biotechnology*, 9 (2021) 791238.

[87] S.J. Hwang, H.S. Choi, H.H. Choi, H.S. Kim, Y.H. Kim, The evaluation of the lower extremity joint moments and muscle force during various high-heel walking, *Key Engineering Materials*, 326 (2006) 755-758.

[88] K. Tokunaga, Y. Nakai, R. Matsumoto, R. Kiyama, M. Kawada, A. Ohwatashi, K. Fukudome, T. Ohshige, T. Maeda, Effect of Foot Progression Angle and Lateral Wedge Insole on a Reduction in Knee Adduction Moment, *J Appl Biomech*, 32 (2016) 454-461.

[89] T. Schmalz, S. Blumentritt, H. Drewitz, M. Freslier, The influence of sole wedges on frontal plane knee kinetics, in isolation and in combination with representative rigid and semi-rigid ankle-foot-orthoses, *Clin Biomech (Bristol)*, 21 (2006) 631-639.

[90] P. Yu, X. Cen, L. Xiang, Q. Mei, A. Wang, Y. Gu, J. Fernandez, Regional plantar forces and surface geometry variations of a chronic ankle instability population described by statistical shape modelling, *Gait Posture*, 106 (2023) 11-17.

[91] Q. Mei, J. Fernandez, W. Fu, N. Feng, Y. Gu, A comparative biomechanical analysis of habitually unshod and shod runners based on a foot morphological difference, *Hum Mov Sci*, 42 (2015) 38-53.

- [92] Z. Lu, D. Sun, B. Kovács, Z. Radák, Y. Gu, Case study: The influence of Achilles tendon rupture on knee joint stress during counter-movement jump - Combining musculoskeletal modeling and finite element analysis, *Heliyon*, 9 (2023) e18410.
- [93] Y. Song, X. Cen, H. Chen, D. Sun, G. Munivrana, K. Bálint, I. Bíró, Y. Gu, The influence of running shoe with different carbon-fiber plate designs on internal foot mechanics: A pilot computational analysis, *J Biomech*, 153 (2023) 111597.
- [94] Y. Lu, Q. Mei, H.-T. Peng, J. Li, C. Wei, Y. Gu, A comparative study on loadings of the lower extremity during deep squat in Asian and Caucasian individuals via OpenSim musculoskeletal modelling, *BioMed research international*, 2020 (2020) 7531719.
- [95] D. Xiong, D. Zhang, Y. Chu, Y. Zhao, X. Zhao, Intuitive human-robot-environment interaction with EMG signals: A review, *IEEE/CAA Journal of Automatica Sinica*, 11 (2024) 1075-1091.
- [96] E. Anselmino, A. Mazzoni, S. Micera, EMG-based prediction of step direction for a better control of lower limb wearable devices, *Computer methods and programs in biomedicine*, 254 (2024) 108305.
- [97] T.J. Dick, K. Tucker, F. Hug, M. Besomi, J.H. van Dieën, R.M. Enoka, T. Besier, R.G. Carson, E.A. Clancy, C. Disselhorst-Klug, Consensus for experimental design in electromyography (CEDE) project: Application of EMG to estimate muscle force, *Journal of Electromyography and Kinesiology*, 79 (2024) 102910.
- [98] D. Farina, R. Merletti, R.M. Enoka, The extraction of neural strategies from the surface EMG: 2004–2024, *Journal of applied physiology*, 138 (2025) 121-135.
- [99] L.I.B. López, F.M. Ferri, J. Zea, Á.L.V. Caraguay, M.E. Benalcázar, CNN-LSTM and post-processing for EMG-based hand gesture recognition, *Intelligent Systems with Applications*, 22 (2024) 200352.
- [100] S. Muceli, R. Merletti, Tutorial. Frequency analysis of the surface EMG signal: Best practices, *Journal of Electromyography and Kinesiology*, 79 (2024) 102937.

- [101] D. Xu, J. Lu, J.S. Baker, G. Fekete, Y. Gu, Temporal kinematic and kinetics differences throughout different landing ways following volleyball spike shots, *Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology*, 236 (2022) 200-208.
- [102] A.C. Jones, R.K. Wilcox, Finite element analysis of the spine: towards a framework of verification, validation and sensitivity analysis, *Medical engineering & physics*, 30 (2008) 1287-1304.
- [103] Q. Zhang, Y. Zhang, J. Huang, E.C. Teo, Y. Gu, Effect of Displacement Degree of Distal Chevron Osteotomy on Metatarsal Stress: A Finite Element Method, *Biology (Basel)*, 11 (2022).
- [104] X. Cen, Y. Song, P. Yu, D. Sun, J. Simon, I. Bíró, Y. Gu, Effects of plantar fascia stiffness on the internal mechanics of idiopathic pes cavus by finite element analysis: Implications for metatarsalgia, *Computer Methods in Biomechanics and Biomedical Engineering*, (2023) 1-9.
- [105] Y. Dabiri, S. Najarian, M.R. Eslami, S. Zahedi, D. Moser, A powered prosthetic knee joint inspired from musculoskeletal system, *Biocybernetics and Biomedical Engineering*, 33 (2013) 118-124.
- [106] A. Saygılı, S. Albayrak, A new computer-based approach for fully automated segmentation of knee meniscus from magnetic resonance images, *Biocybernetics and Biomedical Engineering*, 37 (2017) 432-442.
- [107] X. Cen, Y. Song, D. Sun, I. Bíró, Y. Gu, Applications of Finite Element Modeling in Biomechanical Analysis of Foot Arch Deformation: A Scoping Review, *Journal of biomechanical engineering*, 145 (2023).
- [108] Q. Zhang, D. Sun, M. Wang, V. Tafferner-Gulyás, H. Chen, I. Bíró, Y. Gu, Effect of plantar fascia stiffness on plantar windlass mechanism and arch: Finite element method and dual fluoroscopic imaging system verification, *Medical Engineering & Physics*, 134 (2024) 104259.

- [109] P.O. Bolcos, M.E. Mononen, A. Mohammadi, M. Ebrahimi, M.S. Tanaka, M.A. Samaan, R.B. Souza, X. Li, J.S. Suomalainen, J.S. Jurvelin, J. Töyräs, R.K. Korhonen, Comparison between kinetic and kinetic-kinematic driven knee joint finite element models, *Scientific reports*, 8 (2018) 17351.
- [110] M. Du, J. Sun, Y. Liu, Y. Wang, S. Yan, J. Zeng, K. Zhang, Tibio-Femoral Contact Force Distribution of Knee Before and After Total Knee Arthroplasty: Combined Finite Element and Gait Analysis, *Orthopaedic surgery*, 14 (2022) 1836-1845.
- [111] J.Y. Zhang, J. Wang, D.M. Tian, D.P. Jiang, J.J. Li, Y.C. Hu, Spherical center and rotating platform hinged knee prosthesis: Finite-element model establishment, verification and contact analysis, *Knee*, 27 (2020) 731-739.
- [112] E. Pena, B. Calvo, M. Martinez, D. Palanca, M. Doblaré, Finite element analysis of the effect of meniscal tears and meniscectomies on human knee biomechanics, *Clinical biomechanics*, 20 (2005) 498-507.
- [113] D. Shriram, G. Praveen Kumar, F. Cui, Y.H.D. Lee, K. Subburaj, Evaluating the effects of material properties of artificial meniscal implant in the human knee joint using finite element analysis, *Scientific reports*, 7 (2017) 6011.
- [114] P. Łuczkiwicz, K. Daszkiewicz, W. Witkowski, J. Chróścielewski, W. Zarzycki, Influence of meniscus shape in the cross sectional plane on the knee contact mechanics, *J Biomech*, 48 (2015) 1356-1363.
- [115] A. Erdemir, Open Knee: Open Source Modeling and Simulation in Knee Biomechanics, *The journal of knee surgery*, 29 (2016) 107-116.
- [116] L. Li, L. Yang, K. Zhang, L. Zhu, X. Wang, Q. Jiang, Three-dimensional finite-element analysis of aggravating medial meniscus tears on knee osteoarthritis, *Journal of orthopaedic translation*, 20 (2020) 47-55.
- [117] Y. Song, X. Cen, H. Chen, D. Sun, G. Munivrana, K. Bálint, I. Bíró, Y. Gu, The influence of running shoe with different carbon-fiber plate designs on internal foot mechanics: A pilot computational analysis, *Journal of Biomechanics*, 153 (2023) 111597.

- [118] J. Bingham, G. Li, An optimized image matching method for determining in-vivo TKA kinematics with a dual-orthogonal fluoroscopic imaging system, *Journal of biomechanical engineering*, 128 (2006) 588-595.
- [119] D. Peng, Z. Mao, W. Zhang, J. Yu, S. Zhang, In vivo knee biomechanics during badminton lunges at different distances and different foot positions by using the dual fluoroscopic imaging system, *Frontiers in bioengineering and biotechnology*, 11 (2023) 1320404.
- [120] J.Y. Bae, G.-H. Kim, J.K. Seon, I. Jeon, Finite element study on the anatomic transtibial technique for single-bundle anterior cruciate ligament reconstruction, *Medical & biological engineering & computing*, 54 (2016) 811-820.
- [121] S.-C. Lin, C.P. Chen, S.F. Tang, A.M. Wong, J.-H. Hsieh, W.-P. Chen, Changes in windlass effect in response to different shoe and insole designs during walking, *Gait & posture*, 37 (2013) 235-241.
- [122] G. Li, L.E. DeFrate, H. Sun, T.J. Gill, In vivo elongation of the anterior cruciate ligament and posterior cruciate ligament during knee flexion, *The American journal of sports medicine*, 32 (2004) 1415-1420.
- [123] P. Schober, C. Boer, L.A. Schwarte, Correlation Coefficients: Appropriate Use and Interpretation, *Anesthesia and analgesia*, 126 (2018) 1763-1768.
- [124] L. Benos, D. Stanev, L. Spyrou, K. Moustakas, D.E. Tsaopoulos, A Review on Finite Element Modeling and Simulation of the Anterior Cruciate Ligament Reconstruction, *Frontiers in bioengineering and biotechnology*, 8 (2020) 967.
- [125] B. Shao, J. Xing, B. Zhao, T. Wang, W. Mu, Role of the proximal tibiofibular joint on the biomechanics of the knee joint: A three-dimensional finite element analysis, *Injury*, 53 (2022) 2446-2453.
- [126] K. Kipp, J. Warmenhoven, Applications of regularized regression models in sports biomechanics research, *Sports Biomechanics*, 24 (2025) 723-741.

- [127] S.R. Hamner, S.L. Delp, Muscle contributions to fore-aft and vertical body mass center accelerations over a range of running speeds, *Journal of biomechanics*, 46 (2013) 780-787.
- [128] A. Rajagopal, C.L. Dembia, M.S. DeMers, D.D. Delp, J.L. Hicks, S.L. Delp, Full-body musculoskeletal model for muscle-driven simulation of human gait, *IEEE transactions on biomedical engineering*, 63 (2016) 2068-2079.
- [129] L.A. Kelly, G. Lichtwark, A.G. Cresswell, Active regulation of longitudinal arch compression and recoil during walking and running, *Journal of the Royal Society Interface*, 12 (2015) 20141076.
- [130] S.K. Sarrafian, Functional characteristics of the foot and plantar aponeurosis under tibiotalar loading, *Foot & ankle*, 8 (1987) 4-18.
- [131] A. Gefen, Stress analysis of the standing foot following surgical plantar fascia release, *Journal of biomechanics*, 35 (2002) 629-637.
- [132] A. Erdemir, A.J. Hamel, A.R. Fauth, S.J. Piazza, N.A. Sharkey, Dynamic loading of the plantar aponeurosis in walking, *JBJS*, 86 (2004) 546-552.
- [133] R.E. Carlson, L.L. Fleming, W.C. Hutton, The biomechanical relationship between the tendoachilles, plantar fascia and metatarsophalangeal joint dorsiflexion angle, *Foot & ankle international*, 21 (2000) 18-25.
- [134] J.C. Christensen, M.M. Jennings, Normal and abnormal function of the first ray, *Clinics in podiatric medicine and surgery*, 26 (2009) 355-371, Table of Contents.
- [135] P. Caravaggi, T. Pataky, J.Y. Goulermas, R. Savage, R. Crompton, A dynamic model of the windlass mechanism of the foot: evidence for early stance phase preloading of the plantar aponeurosis, *The Journal of experimental biology*, 212 (2009) 2491-2499.
- [136] N.L. Griffin, B.G. Richmond, Joint orientation and function in great ape and human proximal pedal phalanges, *American journal of physical anthropology*, 141 (2010) 116-123.

- [137] E. Vereecke, K. D'Août, L. Van Elsacker, D. De Clercq, P. Aerts, Functional analysis of the gibbon foot during terrestrial bipedal walking: Plantar pressure distributions and three-dimensional ground reaction forces, *American Journal of Physical Anthropology: The Official Publication of the American Association of Physical Anthropologists*, 128 (2005) 659-669.
- [138] R. Puttaswamaiah, P. Chandran, Degenerative plantar fasciitis: A review of current concepts, *The foot*, 17 (2007) 3-9.
- [139] L.A. Bolgla, T.R. Malone, Plantar fasciitis and the windlass mechanism: a biomechanical link to clinical practice, *Journal of athletic training*, 39 (2004) 77.
- [140] A. Hosseini, S.K. Van de Velde, M. Kozanek, T.J. Gill, A.J. Grodzinsky, H.E. Rubash, G. Li, In-vivo time-dependent articular cartilage contact behavior of the tibiofemoral joint, *Osteoarthritis and cartilage*, 18 (2010) 909-916.
- [141] L.T. Atkins, C. Smithson, D. Grimes, N. Heuer, The influence of sagittal trunk posture on the magnitude and rate of patellofemoral joint stress during stair ascent in asymptomatic females, *Gait & Posture*, 74 (2019) 121-127.
- [142] L. Shu, K. Yamamoto, R. Yoshizaki, J. Yao, T. Sato, N. Sugita, Multiscale finite element musculoskeletal model for intact knee dynamics, *Computers in biology and medicine*, 141 (2022) 105023.
- [143] B.J. McFadyen, D.A. Winter, An integrated biomechanical analysis of normal stair ascent and descent, *Journal of biomechanics*, 21 (1988) 733-744.
- [144] J.-S. Li, T.-Y. Tsai, M.M. Clancy, G. Li, C.L. Lewis, D.T. Felson, Weight loss changed gait kinematics in individuals with obesity and knee pain, *Gait & posture*, 68 (2019) 461-465.
- [145] M.R. Angerame, D.C. Holst, J.M. Jennings, R.D. Komistek, D.A. Dennis, Total knee arthroplasty kinematics, *The Journal of arthroplasty*, 34 (2019) 2502-2510.
- [146] C.A. Frigo, M. Grossi, L. Donno, Loads on the Knee Joint Ligaments during Stair

Climbing, *Applied Sciences*, 13 (2023) 7388.

[147] L. Li, Y. Song, B. Dai, A systematic review of in vivo anterior cruciate ligament loading during static, slow-speed and athletic tasks, *Journal of Science in Sport and Exercise*, 6 (2024) 1-13.

[148] A. Schiavone Panni, S. Cerciello, C. Del Regno, A. Felici, M. Vasso, Patellar resurfacing complications in total knee arthroplasty, *International orthopaedics*, 38 (2014) 313-317.

[149] D. Simon, A. Kleyer, S. Bayat, J. Knitza, L. Valor-Mendez, M. Schweiger, G. Schett, K. Tascilar, A.J. Hueber, Biomechanical stress in the context of competitive sports training triggers enthesitis, *Arthritis research & therapy*, 23 (2021) 172.

[150] E. Gracey, A. Burssens, I. Cambré, G. Schett, R. Lories, I.B. McInnes, H. Asahara, D. Elewaut, Tendon and ligament mechanical loading in the pathogenesis of inflammatory arthritis, *Nature reviews. Rheumatology*, 16 (2020) 193-207.

[151] S.J. Pearson, S.R. Hussain, Region-specific tendon properties and patellar tendinopathy: a wider understanding, *Sports medicine (Auckland, N.Z.)*, 44 (2014) 1101-1112.

[152] M. Lavagnino, S.P. Arnoczky, N. Elvin, J. Dodds, Patellar tendon strain is increased at the site of the jumper's knee lesion during knee flexion and tendon loading: results and cadaveric testing of a computational model, *The American journal of sports medicine*, 36 (2008) 2110-2118.

[153] K. Wang, S.H. Hosseinnajad, A. Jabran, V. Baltzopoulos, L. Ren, Z. Qian, A biomechanical analysis of 3D stress and strain patterns in patellar tendon during knee flexion, *International journal for numerical methods in biomedical engineering*, 36 (2020) e3379.

[154] K. Park, G. Salem, C.M. Powers, Persons with patellar tendinopathy exhibit anatomical features that alter knee extensor mechanics: A comparison of persons with and without symptoms, *J Biomech*, 144 (2022) 111343.

- [155] K. Park, J.H. Keyak, C.M. Powers, The influence of isolated femur and tibia rotations on patellar tendon stress: A sensitivity analysis using finite element analysis, *Journal of orthopaedic research : official publication of the Orthopaedic Research Society*, 41 (2023) 271-277.
- [156] S. Saadat, M.L. Stephenson, J.C. Gillette, Entry angle during jump landing changes biomechanical risk factors for ACL injury, *Sports Biomech*, (2022) 1-13.
- [157] R. Ramirez-Campillo, A. García-Hermoso, J. Moran, H. Chaabene, Y. Negra, A.T. Scanlan, The effects of plyometric jump training on physical fitness attributes in basketball players: A meta-analysis, *Journal of sport and health science*, 11 (2022) 656-670.

ABBREVIATION

STA: Soft Tissue Artifact	ALB: Anterior-lateral Bundle
CT: Computed Tomography	PMB: Posterior-medial bundle
MRI: Magnetic Resonance Imaging	MCL: Medial Collateral Ligament
DFIS: Dual-plane Fluoroscopy Imaging System	LCL: Lateral Collateral Ligament
6-DOF: six-degrees-of-freedom	PT: Patellar Tendon
FE: Finite Element	ACL: Anterior Cruciate Ligament
FEA: Finite Element Analysis	PCL: Posterior Cruciate Ligament
FEM: Finite Element Method	PT: Patellar Tendon
MLA: Medial Longitudinal Arch	EMG: Electromyography
LLA: Lateral Longitudinal Arch	AHI: Arch Height Index
GRF: Ground Reaction Forces	FPI: Foot Posture Index
ROM: Range of Motion	GC: Gastrocnemius
PF: Plantar Fascia	SL: Soleus

MTPJ: Metatarsophalangeal Joint

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ACKNOWLEDGEMENTS

This doctoral journey has been a remarkable mix of challenges and rewards, and I am deeply humbled by the unwavering support and encouragement I have received along the way. As it now draws to a close, my heart is filled with both reluctance to bid farewell and profound nostalgia.

First and foremost, I extend my deepest thanks to my esteemed supervisors, Prof. Dr. István Bíró and Prof. Dr. Yaodong Gu, for their invaluable guidance, insightful contributions, and boundless patience. Their expertise and dedication have been instrumental in shaping the course of this research.

I also want to express my sincere appreciation to the staff and faculty at the Doctoral School on Safety and Security Sciences, Óbuda University (OE), the Faculty of Engineering, University of Szeged (SZTE), and the Faculty of Sports Science, Ningbo University (NBU), for providing me with a conducive research environment and the necessary resources to conduct this study. Moreover, I extend my heartfelt thanks to the Stipendium Hungaricum Programme, Tempus Public Foundation, and China Scholarship Council (CSC) for their generous financial support, which has been instrumental in making this journey possible.

To my family, I am eternally thankful for your unwavering love, encouragement, and unyielding belief in me. Your constant support and understanding throughout this entire journey have been an unshakable source of strength and motivation.

Last but not least, I want to express my deep appreciation to the countless researchers, scholars, and academics whose contributions and insights have profoundly shaped my thinking and laid a solid foundation for this work. I feel truly indebted to the broader academic community for their collective efforts in advancing knowledge in the field of **Biomechanics**.

I am profoundly grateful to each one of you for being part of this significant milestone in my academic journey.

Qiaolin Zhang