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From uterine mechanics to motor control: a personal perspective on biomechanics across scales

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Abstract

This paper offers a personal perspective on a lifetime of research in biomechanics, organized around the mechanical principles that connect muscle, joint, tissue, limb, and whole-body function. The narrative begins with uterine biomechanics during labor, where in-vivo strain measurements and shell modeling were used to examine contraction patterns, deformation, pacemaker activity, and the transition from near-isotropic to anisotropic behavior. It then turns to joint mechanics, including hip incongruity, acetabular contact pressure, implant fixation, knee arthroplasty, and the gliding index, followed by studies of rehabilitation biomechanics at the Loewenstein Rehabilitation Center. These sections examine gait recovery, stepping and jumping, postural control, bilateral force-platform measurements, and mathematical modeling of body sway across neurological, orthopedic and amputee populations. Subsequent sections address articular cartilage mechanics and cartilage tissue engineering, surgical modifications of the musculoskeletal system, impact loading and fatigue in running, mechanical impedance as a task-dependent expression of motor control, and Functional Electrical Stimulation (FES) for rehabilitation. The FES work includes muscle recruitment, fatigue monitoring using ^{31}P magnetic resonance spectroscopy, fatigue modeling, EMG-based monitoring, hybrid activation and bioelectric field modeling. The final sections consider bone quality, osteoporosis, metastatic weakening, vertebral fragility, and biomechanical indeterminacy. Across these topics, the central theme is that biomechanics is most powerful when experiment, modeling, and clinical observation are integrated. The concluding section reflects on how indeterminacy, redundancy, adaptation, and structure-function relations unify seemingly diverse problems in biomechanics.

Key Words: uterine muscle, joint mechanics, impact loading, tissue engineering, muscle stimulation, muscle fatigue, mechanical impedance, rehabilitation biomechanics, fracture risk of bone, biomechanical indeterminacy.

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My PhD on the uterine muscle

I received a BSc in Aerospace Engineering from the Technion (1967) and continued to an MSc in mechanics, where I was influenced by Professor Zvi Karni to pursue biomechanics. I became the first PhD student in the Technion's Biomedical Engineering (BME) Unit. My doctoral thesis, "Deformation Analysis of the Uterine Muscle Activity during Labor and Delivery",[1] supervised by Karni and Polishuk, summarizes my work on uterine biomechanics.

Biomechanics of the uterus

At the time, research focused mainly on hormonal and physiological aspects of uterine activity,[2] with limited attention to mechanical behavior. My work addressed this gap by combining modeling with direct strain measurements.

Modeling and measurement of uterine activity

The uterus during labor was modeled as a thin, axisymmetric muscular shell with two interacting fiber families (longitudinal and circular).[3] Contraction generates internal pressure and deformation, transforming the uterus from a pear-like toward a spherical shape.[1,3,4]

Strain was measured in vivo, to provide input data for the model, using high-extension strain gauges. The measurements enabled non-invasive assessment of contraction patterns, pacemaker behavior, and fetal activity. These measurements also supported clinical monitoring of fetal well-being, where reduced movement may indicate distress.[5]

The key findings indicated that the uterine contractions originate from a dominant fundal pacemaker and propagate downward at ~5.7 cm/s). The strain patterns show extension in the fundus and contraction in the lower uterus, with a

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transition zone between them.[5] Measured time delays support a coordinated single-dominant pacemaker rather than dual sources.[6] Interestingly, early labor exhibits near-isotropic behavior and later stages become increasingly anisotropic as fiber groups act more independently.[7,8] This is tested using strain rosettes, which measure deformation in multiple directions to compare predicted and actual strains. The work on uterine activity in labor established a mechanical framework for understanding uterine function by integrating theory and experiment. Apart from its journal publications, our work was recognized through an invitation to author Chapter 10 in the "Handbook of Bioengineering", a foundational volume in the field.[9]

Hip joint mechanics and contact behavior (University of the Witwatersrand)

In 1976, I joined Prof. Louis Solomon's Department of Orthopaedic Surgery at the University of the Witwatersrand in Johannesburg as a research fellow, following my PhD. His work focused on mechanical factors in osteoarthritis, particularly in the hip.[10] My role was to investigate biomechanical stress distribution and its relation to disease. A joint appointment with Mechanical Engineering enabled interdisciplinary collaboration, laboratory access, and supervision of student projects.

Joint geometry and contact mechanics

We studied hip joint incongruity, where shifting contact areas enhance synovial fluid circulation.[11] Prior observations of femoral head curvature variation lacked a unifying model.[12] Applying surface geometry methods, I showed that the femoral head can be modeled as a solid of revolution with an axis offset from the femoral neck. This enabled analytical load-distribution modeling and improved anatomical representation for prosthetic design.[13]

Contact pressure measurements

We developed a method to measure local acetabular cartilage contact pressures in vitro, using implanted transducers, inserted through the subchondral bone plate and positioned immediately beneath and in contact with the cartilage. A key technical challenge was determining the precise direction and depth of drilling through the irregular bony geometry of the ilium, ischium, and pubis.[14] While pressures varied across specimens, consistent patterns emerged: peak stresses occurred anteriorly and posteriorly, whereas the acetabular zenith was relatively unloaded—consistent with prior findings.[15] Notably, osteoarthritic degeneration often occurs in this less-loaded region, supporting the role of atypical mechanical factors.[10]

Hip implant studies

Collaborative work between orthopaedic surgery and mechanical engineering led to two studies: i) Charnley prostheses: Strain-gauge measurements showed that stress distributions are highly sensitive to implant orientation, with implications for fatigue failure under cyclic loading;[16] ii) femoral-neck fixation: Optimal screw configurations, using Muller cancellous screws, depend on bone quality and geometry—three screws (triangular) for osteoporotic bone, four screws for stronger bone.[17]
Several senior student projects contributed substantially to these studies, fostering early engagement in biomechanics.

Knee arthroplasty and the gliding index

A few years later, we extended the work on joints to the knee and addressed the balance between kinematics and geometry in prosthesis design.[18] We introduced the gliding index, representing tangential forces contributing to wear and loosening.[19] Using gait data from the literature[20] and additional self-measurements, we developed optimized surface geometries.[21]
This work was carried out with Eric Benaim, a brilliant mathematician, at the Technion in the mid 1980's. Eric joined my laboratory after completing engineering studies at l'École Polytechnique des Mines in Paris. He later continued with me as a PhD student in cartilage biomechanics. Tragically, Eric passed away prematurely just a few years ago.

Rehabilitation biomechanics at the Loewenstein Rehabilitation Center, Raanana

In 1979, after three years in Johannesburg, I returned to Israel as Head of the Biomechanics Laboratory at the Loewenstein Rehabilitation Center in Raanana. My research focused on gait, posture, joint mechanics, and functional electrical stimulation.
My principal collaborators were Professors Theodore Najenson and Zeev Susak, clinicians deeply committed to biomechanics research. Najenson, the hospital's medical director, wanted to introduce engineering methods into rehabilitation practice and became both colleague and friend. Susak, Head of Orthopaedic Rehabilitation, was also a close friend and valued partner in discussion.
It should be noted that comprehensive analysis of human locomotion relies on multi-parameter measurements that integrate kinematic variables (e.g., linear and angular displacement, velocity, and acceleration), kinetic data (e.g., forces and torques), and muscle activity assessed via Electromyography (EMG). Together, these generate a substantial volume of information, which complicates the development of clinically practical and efficient gait evaluation methods. Furthermore, general gait analysis is inherently affected by solution uncertainties, arising from indeterminacies where the number of unknowns exceeds the available equations.
Our approach was thus to investigate whether selected locomotor abnormalities could be characterized through a more focused approach. Accordingly, we concentrated on the measurement and analysis of gait pathologies within a laboratory setting, with particular emphasis on the lower limb—examining stride kinematics, foot–ground reaction forces, and the dynamic aspects of standing posture.
Our gait laboratory enabled systematic studies of gait and postural stability. We developed an apparatus for measuring time-distance parameters of the stride and used it to follow post-stroke gait recovery. The measurements indi-

cated that velocity, stride length, symmetry, double-support time, and stance time improved and became less variable. A logarithmic recovery pattern helped identify rehabilitation plateaus, and walking speed served as a useful global measure.[22,23,24]

The same walkway was used to assess walking after internal fixation of intertrochanteric hip fractures. Persistent limitations appeared to reflect the lasting impact of hip fractures in older adults rather than implant-specific differences.[25] We also quantified outcomes after equinus correction in children with cerebral palsy, finding improvements in support time, speed, and stride length beyond those expected from growth alone.[26]

Another line of work examined body dynamics using forceplates. In below-knee amputees, we assessed prosthetic alignment and stump-socket stability through phase durations, peak forces, impulses, and force rates. Several variables were clinically useful, and changes in the antero-posterior force curve appeared to reveal specific prosthetic malfunctions.[27]

This research was part of a long collaboration with Professor Rami Seliktar of Drexel University on rehabilitation biomechanics. We shared together the supervision of graduate students, attended conferences together, and published numerous joint papers. Apart from occasional mutual visits, Rami and I exchanged two extended sabbatical visits at Technion and Drexel, respectively.

Stepping and jumping down motion

Stepping and jumping down generates high knee and ankle loads. With the knee highly flexed, its forward position increases the moment arm of the vertical reaction force and therefore the resulting flexing knee moment.[28] Using a quasi-static model based on ground-reaction forces and joint-angle data, we showed that forward steps and jumps produce a backward braking force that reduces the dorsiflexing ankle moment but increases knee-load moments. In backward stepping, the same force increases ankle loading while reducing knee loading. This may explain why some patients with knee pain prefer backward stepping. Ralph Nisell of the Karolinska Institute contributed actively to this work while visiting my lab.[29]

The dynamic nature of standing

Quiet standing is not a static state. It resembles an inverted pendulum that must be continuously stabilized by coordinated activity in the lower limbs and trunk. The resulting body sway reflects the integrated function of the musculoskeletal system and the visual, vestibular, and proprioceptive control loops. Aging, disease, trauma, or limb loss may disturb any of these inputs and thereby compromise upright stability.[30]

At the Loewenstein Biomechanics Laboratory, we developed a bilateral force-platform approach to quantify standing balance. Instead of measuring only the resultant ground reaction force beneath the whole body, two adjacent force platforms recorded the reactive forces under each foot separately. This made it possible to characterize the contribution of each limb to postural control, an important distinction in asymmetric clinical populations.[31]

Bilateral force-platform measures

From the force traces, we extracted waveform timing and amplitude, the vectorial sequence of horizontal force oscillations in both feet, and the Relative Sequence of Tangential Vectors (RSTV; Figure 1). Weight-Bearing Imbalance (WBI) quantified the verti-

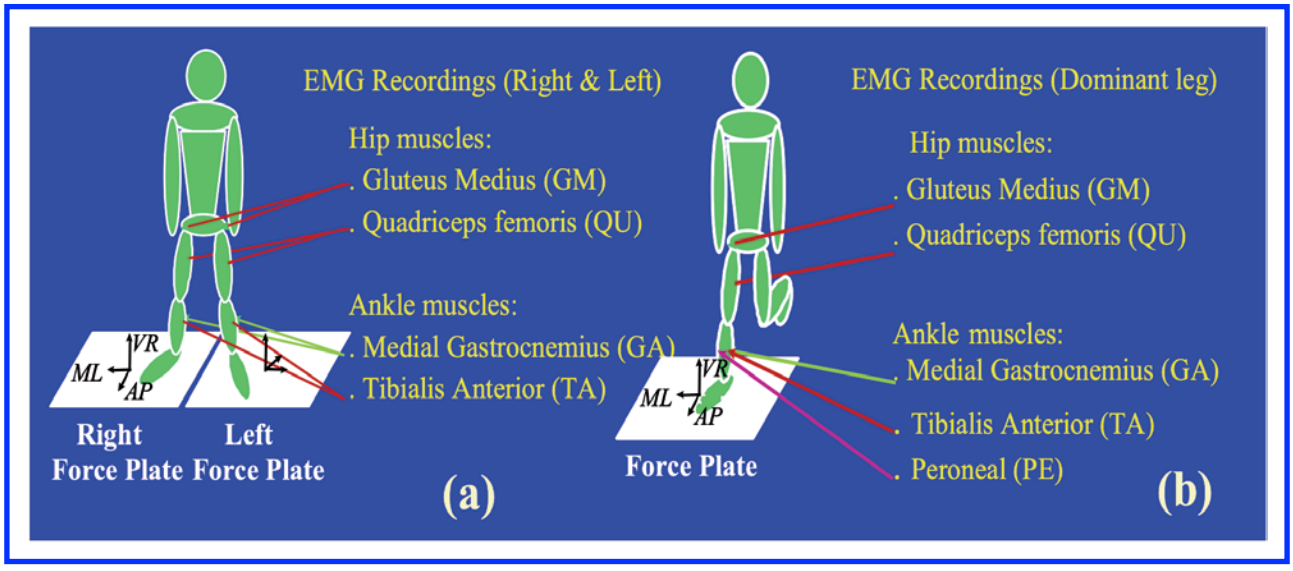

Figure 1*. Doble-supported* versus *single-supported standing. Reactive forces measured are represented by the VR (vertical), AP (anterior-posterior) and ML (medio-lateral) components. EMG recordings include the Gluteus Medius (GM), Quadriceps Femoris (QU), Medial Gastroctenius (GA), Tibilias Anterior (TA), and Peroneal PE) muscles.*

cal load difference between the limbs. Two horizontal-plane measures were also defined: Sway Activity (SA), the vector sum of the absolute horizontal force components and a proxy for total stabilizing muscular activity; and Asymmetry (ASYM), the difference in activity between the two legs.

In able-bodied participants, WBI averaged about 5.4% of body weight, SA was approximately 0.5% of body weight, and ASYM was about 0.07% of body weight. Although the bilateral force traces were synchronous, RSTV patterns varied across individuals, showing that each limb may use a distinct stabilizing strategy. This supported the view that standing balance should be analyzed segmentally rather than only through whole-body resultant forces.[31]

Clinical applications

The method was then applied in collaboration with the Loewenstein rehabilitation departments. In post-CVA hemiplegic patients, the bilateral forces remained largely synchronous but became markedly asymmetrical. SA was higher than in controls, averaging about 0.74% of body weight, and ASYM rose to about 0.27%. These findings indicated that stroke patients required greater muscular activity to maintain equilibrium and relied disproportionately on the sound limb. RSTV, lower SA, and lower asymmetry were also associated with better locomotor outcomes, suggesting their value for prognosis and rehabilitation monitoring.[32,33]

In Craniocerebral Injury patients (CCI), bilateral force measurements likewise showed synchronous but often asymmetric patterns. SA and ASYM tended to increase with the severity of neurobehavioral impairment, indicating that postural sway measures could reflect the extent of neurological disturbance.[34]

Studies of unilateral below-knee amputees showed significantly reduced stability compared with able-bodied controls. Sway activity, particularly in the anteroposterior direction, was greater in the contralateral limb than in the prosthetic limb, consistent with proprioceptive loss and altered loading strategies after amputation. In some participants, sway decreased toward the end of rehabilitation, suggesting compensation and adaptation. Prosthetic alignment was also clinically relevant: varus-valgus alignment changes were associated with discomfort and increased sway instability.[35]

A comparative study of hemiparetic subjects, below-knee amputees, and healthy adults showed persistent weight-bearing asymmetry in both clinical groups, with reduced loading of the affected or prosthetic limb. These findings reinforced the importance of bilateral force measurement for identifying postural deficits and guiding rehabilitation.[36]

Postural sway in subjects with recurrent ankle sprains

Instrumented balance testing was also used to examine people with recurrent ankle sprains. During single-leg standing, no clear static balance deficit was detected in the sprained ankle. Across participants, however, closing their eyes during the test increased reaction forces and reduced stability. These results suggested that chronic ankle instability does not necessarily imply a measurable proprioceptive deficit during quiet stance and supported earlier findings that peroneal muscle latency did not differ significantly between healthy subjects and individuals with recurrent sprains.[37,38]

Modeling postural sway

With graduate student Oron Levin, we introduced mathematical modeling of body sway. The first objective was to estimate the trajectory of the body Center of Gravity (CG) from bilateral ground reaction forces and centers of pressure. This required a three-dimensional, four-joint, five-segment model and an iterative algorithm based on the equations of motion and angular momentum. The model provided quantitative insight into sway, gait, and the feedback mechanisms that integrate visual, vestibular, and proprioceptive information.[39,40]

Single-stance standing involves the continuous postural orientation and alignment of multiple body segments, accompanied by extensive and readily measurable Ground Reaction Force (GRF) activity. The orientation of the trunk is likely one of the most important controlled variables, as it determines the relative positioning of the limbs. Therefore, it would be useful to develop a model that uses GRFs as inputs to evaluate the position and orientation of the trunk and other body segments in response to joint displacements. This problem is analogous to determining the position of a robotic manipulator's end-effector from the positions of its joints.

Compared to two-leg standing, the problem in single-leg standing becomes less redundant and mechanically more determinate. A three-segment open-chain model with four rotational coordinates, two at the ankle and two at the hip, was developed using Denavit-Hartenberg notation.[41] The CG trajectory was estimated by combining angular momentum principles with direct integration of the equations of motion. Joint torque and angular displacement sequences showed that power delivered at one joint could be absorbed by adjacent muscles, suggesting a strategy for stabilizing the CG while minimizing muscular effort.[42]

We also examined how selective fatiguing of the tibialis anterior and peroneal muscles affects single-leg postural control. The induced muscle imbalance increased sway and strengthened the correlation between muscle activity and sway parameters, making the system more synchronized. These findings have relevance for muscle impairment and for cases in which electrical stimulation is used to augment muscle activity.[43]

In conclusion, across healthy, neurological, orthopedic, amputee, infant, and modeling studies, the central message was consistent: standing balance is dynamic, bilateral, and limb specific. Measuring each limb separately reveals asymmetries, compensatory strategies, and control mechanisms that whole-body measurements can obscure. This approach offers clinically useful markers for assessment, prognosis, rehabilitation planning, prosthetic alignment, and the mathematical modeling of human postural control.[30]

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Articular cartilage

In 1982, I joined the Technion while continuing to lead the Biomechanics Laboratory at the Loewenstein Rehabilitation Center and began collaborating with Alice Maroudas on links between cartilage mechanics and composition.

We studied articular cartilage under unconfined compression, capturing both fluid flow and lateral deformation. Contrary to the assumed axisymmetry,[44] we showed that anisotropy produces direction-dependent radial strain: circular samples became elliptical, with the major axis perpendicular to collagen-fiber orientation.[45,46] Our analysis incorporated concurrent changes in shape, area, and thickness.

We found that instantaneous response is governed by the collagen network, while creep is dominated by fluid flow, with anisotropic stiffness consistent with tensile data.[46-48] The combined experimental–theoretical framework enabled estimation of osmotic pressure and hydraulic permeability as functions of proteoglycan content.[49,50]

These results established unconfined compression as a sensitive tool for distinguishing healthy from degenerated cartilage,[51-53] and for quantifying fluid transport and matrix organization, including two-compartment models of cartilage water.[54,55]

Cartilage tissue engineering

Our cartilage tissue engineering project arose from a broader joint Technion–Hopkins collaboration in biomedical engineering. The collaboration was conceived during a September 2001 visit to Johns Hopkins University by a small Technion BME delegation, of which I was a member. Notably, this collaboration began under circumstances that underscored the unpredictability of global events yet led to a highly productive scientific partnership (Figure 2). At the same time, cartilage tissue engineering was emerging as both a promising therapeutic strategy and an important research field, and I was pleased to collaborate with Professor Jennifer Elisseeff, whose expertise contributed significantly to the project.

One year later, Dr. Dror Seliktar—incidentally, the son of Professor Rami Seliktar—joined the Technion BME Department after training with Robert Nerem in Atlanta and Jeffrey Hubbell in Zurich. His arrival significantly strengthened our department's tissue engineering activity. A laboratory was soon established, joint students joined the effort, and I was pleased to the inclusion of Dror in the recently approved research grant with Jennifer as a co-principal investigator.

Using PEG-based hydrogels—combining tunable mechanics with biological cues—we examined matrix deposition under dynamic loading.[56,57] Our outstanding group

Figure 2. *With hosts at John Hopkins University.*

of Technion graduate students made significant, indeed leading, contributions to the advancement of the cartilage tissue engineering project. Their work spanned the full spectrum of the effort, from observing cellular behavior to biomechanical modeling and the integration of these insights through material chemistry and programming.

Dynamic loading enhanced cartilage formation, though fibrinogen immobilization offered no added benefit.[58] Adult stem cells responded directly to loading, whereas embryonic cells required prior biochemical conditioning.[59] Load-sensing bioreactors enabled real-time mechanical monitoring and identification of optimal stimulation stages.[60]

Shifting to scaffold design, less restrictive PEG-albumin gels promoted superior matrix accumulation versus PEG-fibrinogen.[61] This was explained by stiffness-dependent cell strain by using a Finite Element Analysis (FEA)[62] and enhanced mechanosensing via adhesive motifs.[63] For the cells to feel the mechanical "pull", biological motifs must be incorporated. In 3D, scaffold stiffness—not ligand density—governed cell spreading, enabling control of morphology through cross-linking. This is crucial for engineering tissues like muscle or nerves where the direction and shape of the cells are as important as the material itself.[64]

Surgical modifications

Between 1984 and 2000, we used biomechanical simulation and experiments to study how surgical or structural changes—joint immobilization, tendon transfer, and prosthetic replacement—affect internal forces, joint mechanics, and function.

Ankle–foot immobilization

Limiting ankle–subtalar motion significantly altered gait, reducing push-off and advancing load transfer earlier in stance. Even partial restriction impaired propulsion, highlighting the ankle–foot complex as critical for efficient walking.[65]

Distal tendon transfer

An elbow model comparing transfers of the biceps and brachialis showed functional trade-offs: biceps transfer better maintained supination, while brachialis transfer produced more consistent flexion torque. This demonstrated the value of simulation in optimizing surgical routing choices.[66]

Forearm amputation and prosthesis

Modeling showed that prosthetic use increases muscle forces, elbow moments, and joint reaction loads, indicating system-wide biomechanical effects. Findings emphasized minimizing prosthesis mass, placing weight proximally, and evaluating internal loads alongside function.[67]

Impact loading and fatigue in running

Foot-strike impacts during running generate shock waves that impose substantial stress on bone, cartilage, and soft tissues, contributing to overuse injuries.[68] Two complementary approaches have been used to understand how the leg attenuates these forces.

A modeling approach characterized the leg as a two-degree-of-freedom viscoelastic system using force-platform and acceleration data. This analysis showed that damping plays a greater role than stiffness in reducing peak accelerations, indicating that the leg functions primarily as a shock absorber.[69]

A behavioral perspective demonstrated that movement strategies—including joint flexion, coordinated multi-joint motion, muscle pre-activation, and surface interaction—actively modulate impact attenuation.[70]

Effects of fatigue on impact mechanics

In 1997, I invited Professor Arkady Voloshin from Lehigh University to spend a sabbatical in my Technion laboratory. During his visit, we collaborated on a study examining the effects of fatigue on musculoskeletal overload and injury in long-distance running. Professor Voloshin had extensive experience in the use of lightweight accelerometers for externally measuring shock loads and wave attenuation in the human skeleton.[71]

Studies were performed on healthy individuals running at the anaerobic threshold, and instrumented with tibial accelerometry, EMG, and respiratory measures. Fatigue—quantified via ventilatory markers and end-tidal CO□[72]—was associated with a progressive increase in impact acceleration, indicating reduced shock-absorbing capability.[73]

Notably, standard EMG measures showed little correlation with fatigue, suggesting that the observed deterioration reflects metabolic rather than peripheral muscular fatigue, likely impairing coordination and synergistic muscle function.[73,74] This reduction in coordinated control increases injury risk.[75,76]

With fatigue, shock transmissibility along the skeleton increases, as shown by tibia–sacrum measurements, helping explain site-specific injury susceptibility.[77]

Muscle activity and imbalance

Fatigue alters muscle activation patterns. Tibialis anterior activity (iEMG and MPF) decreases, indicating reduced function, whereas gastrocnemius activity is maintained or increases. This dorsiflexor–plantarflexor imbalance, combined with higher impact loads, may elevate tibial stress and fracture risk.[74]

Kinematic adaptations to fatigue

Progressive fatigue leads to systematic biomechanical changes: reduced stride rate, increased knee extension and reduced flexion, and greater hip vertical excursion. These changes diminish shock absorption capacity and increase mechanical loading.[78]

Downhill running and eccentric fatigue

Downhill running produces greater shock transmission than level running due to eccentric muscle demands. With fatigue—especially in the quadriceps—shock attenuation

declines further, and kinematic adaptations increase loading and injury risk.[79]

Metabolic modulation of fatigue

Sodium bicarbonate supplementation improves buffering capacity, increasing peak force and delaying fatigue, although its direct effect on shock attenuation remains untested.[80]

Mechanical impedance in motor control

In 2013, I spent the spring semester in the Department of Biomedical Engineering at National Cheng Kung University in Tainan, Taiwan, as the guest of Prof. JJ Jason Chen. It was a special honor to participate in the teaching program for his students. During this period, I also worked on an invited paper titled "Mechanical Impedance and Its Relations to Motor Control, Limb Dynamics, and Motion Biomechanics," for the Journal of Medical and Biological Engineering, where Prof. Chen served as Chief Editor.[81] Figure 3 shows distinguished guests from Taiwan during a reciprocating visit to Israel.

Mechanical impedance links neural control, musculoskeletal dynamics, and movement biomechanics. Hogan defined impedance control as regulation of the force–motion relationship to ensure stability, showing that coactivation can tune stiffness and damping.[82] Extending this framework, impedance is best viewed as a task-dependent, time-varying outcome of muscle activation, passive tissues, and limb dynamics, shaping stability, energy transfer, and injury risk.[81]

Injury mechanics and perturbations

Injury risk arises when required limb impedance during impact exceeds neuromuscular capacity. For example, ankle sprain occurs when inversion velocity outpaces peroneal reflex response.[38]

Using an experimental system to track ankle inversion kinematics, and modelling ankle dynamics, we quantified joint stiffness, damping, and inertia. Passive properties increase with inversion and contribute to stability and energy dissipation but are insufficient alone—muscle activation typically occurs too late.[83]

EMG and subsequent work show that pre-activation and

Figure 3. *With distinguished guests from Taiwan at the Bahai garden balcony, overlooking Haifa Bay.*

co-contraction are critical, indicating that injury risk depends not only on reflex delay but also on baseline muscle state.[84]

Locomotion

We examined whether joint impedance during hopping is constant or dynamically modulated. Using inverse dynamics and system identification,[85] we found that stiffness and damping vary throughout stance: The ankle behaves approximately as a quasi-linear spring-damper; the knee shows strong phase-dependent stiffness modulation; and the hip is highly variable, reflecting greater neural control. This modulation supports compliance at landing and stiffness during push-off.

Fatigue further alters impedance,[86] demonstrating that it cannot be constant. Impaired regulation leads to altered impact mechanics and increased injury risk.

Upper-limb control and manipulation

In upper-limb tasks, impedance—especially damping—stabilizes interactions with unstable objects. Roth et al. showed that when carrying a liquid-filled cup, subjects reduce oscillations by increasing damping rather than stiffness.[87] This paper was finalized while I was spending a sabbatical semester at Drexel University, collaborating with Rami Seliktar on research focusing on the mechanical impedance of the upper limb.

This strategy minimizes spill risk without excessive stiffness, highlighting impedance as an active control mechanism rather than a passive property. Tasks involving internal dynamics prioritize damping, while position-control tasks rely more on stiffness.

Subsequent work used EMG-based modeling to estimate muscle contributions to joint impedance during cycling, supporting its interpretation as a quantifiable, task-specific output of neuromuscular control.[88]

A we noticed, mechanical impedance emerges as a central, adaptable expression of motor control across injury mechanics, locomotion, and manipulation. It is not fixed, but continuously regulated to meet task demands, balancing stability, efficiency, and safety.

FES activation of muscles

In the early 1980's, the Functional Electrical Stimulation (FES) project for restoring supported standing and walking in individuals with upper motor neuron Spinal Cord Injury (SCI) patients was initiated in the Biomechanics Laboratory at the Loewenstein Rehabilitation Center, following a brief visit by Professor Daniel Graupe, a long-time faculty member at the University of Illinois at Chicago. Early work of this collaboration focused on training protocols and biomechanical evaluation of gait (weight-bearing, stepping speed, stride timing), demonstrating measurable functional improvement.[89]

Subsequent studies quantified the physiological cost of FES-assisted ambulation, showing high metabolic demand via elevated heart rate, oxygen consumption, and lactate.[90]

System development

A percutaneous, microprocessor-controlled FES system was developed alongside training. It integrated: a six-channel constant-current stimulator and an instrumented walker triggering stimulation. This design improved coordination and user engagement in task-oriented rehabilitation.[91] Key adjustable parameters included intensity (0–400 mA), frequency (10–50 Hz), and pulse width (100–300 μs).

Yoel Minzly, our laboratory technician, was responsible for the development of hardware systems, including a microprocessor-controlled stimulator, artifact suppressor, and related apparatus. He demonstrated exceptional technical skill and capability.

Muscle recruitment and fatigue

Levy et al. characterized muscle recruitment as sigmoidal and fatigue as a double-phase force decay.[92] Using ^{31}P-MRS, metabolic changes during FES revealed rapid phosphocreatine depletion, acidification, and early fatigue.[93]

Mark Levy, MD, joined my lab after immigrating to Israel from Venezuela to pursue a PhD in Biomechanics, focusing on Functional Electrical Stimulation (FES). His doctoral research concentrated on fatigue induced by FES, addressing its mechanical, myoelectric, and metabolic aspects.

Fatigue modeling

In the early 1990's, Dr. Yohanan Giat joined my lab in the Technion for a post-doc fellowship, after completing his PhD at the University of Maryland with Professors William S. Levine and Jigien Chen. Together, Yohanan and I developed a Huxley-type musculo-tendon model incorporating a fatigue parameter derived from intracellular pH and enabling accurate prediction of force decay.[94] This model was later extended to include intermittent stimulation by integrating recovery dynamics via recursive pH functions, optimizing stimulation duty cycles.[95,96]

EMG monitoring in the presence of FES

To extract EMG during stimulation, an artifact-suppression circuit was developed.[97] Analysis revealed a nonlinear EMG–force relationships and that the EMG preceded force during fatigue. These findings support EMG-based real-time fatigue monitoring.[98] Further work showed increasing hamstring co-activation during fatigue, affecting torque distribution.[99]

Therapeutic Electrical Stimulation (TES)

Low-intensity FES (TES) was applied in children with cerebral palsy, improving gait parameters and showing bilateral effects.[100] However, long-term training was found more effective than real-time assistance.[101]

Hybrid activation

In cases of partial deficit in muscle activation capacity, force can be augmented through hybrid activation—combining low-intensity Electrical Stimulation (ES) with voluntary effort. However, in this mode, the respective

contributions of voluntary and stimulation-induced torques to the total output are not directly measurable. To address this, a computational approach was developed to separate these components during combined activation.

Eran Langzam joined the project as my PhD student, with Professor Yael Nemirovsky (Technion, Electrical Engineering) as his co-supervisor. An experimental setup was designed in which the Tibialis Anterior (TA) muscle underwent low-intensity ES to produce isometric contractions under three conditions: i) ES alone, ii) voluntary activation alone, and iii) hybrid activation. Ankle torque and TA EMG were monitored in real time. A computational algorithm was developed to estimate the voluntary torque component based on EMG filtering and pre-calibrated relationships between EMG and voluntary torque.[102]

Results showed that, for a given hybrid torque, the stimulation-induced and voluntary torque components exhibit a linear inverse relationship. Furthermore, using a defined enhancement ratio, regions of facilitated performance were identified, where stimulation efficiency was higher under hybrid than isolated conditions.[103]

In a follow-up experiment, subjects were asked to generate a gait-like dynamic torque pattern with the TA muscle, guided by a visual display.[104] The findings quantified how decreases in voluntary torque must be compensated by increased ES contribution. The method also enabled determination of the ES intensity profile required to achieve a target torque output, providing a foundation for the development of adaptive rehabilitation systems based on hybrid muscle activation.

Bioelectric field modeling

Leonid Livshitz joined my laboratory at the Technion as a graduate student shortly after immigrating to Israel from Moscow. With his strong background in mathematics and physics, he was exceptionally well suited to work on modeling Functional Electrical Stimulation (FES) at multiple scales—from the tissue level, through layered media, down to the cellular and axonal levels. Professor Pinchas Einziger of the Technion's Department of Electrical Engineering, an expert in bioelectromagnetics and nonlinear wave phenomena, joined me as Leonid's co-supervisor.

First, we introduced an image-series formulation for layered tissue—including skin, fat, and muscle—that enabled accurate prediction of Isometric Recruitment Curves (IRCs) and Current-Density (CD) patterns.[105] In the next stage, we extended the finite-electrode model to multiple electrodes and analyzed how electrode size, shape, and arrangement modify current distribution. This work revealed that recruitment depends more strongly on electrode configuration than on electrode size.[106, 107]

We then presented a general formulation based on a rigorous mathematical derivation of Green's functions, providing a mathematical foundation for modeling electrical fields in stratified biological media.[108]

Subsequently, we developed cellular models that enabled precise prediction of transmembrane potentials under arbitrary fields.[109] Further extensions to myelinated axons modeled conduction dynamics and the effects of demyelination.[110]

Relationship between bone quality and fracture risk

This work originated with Dr. J. Y. Margulies, who, during his orthopaedic residency, sought professional collaboration with our biomechanics laboratory. Motivated by his interest in bone fractures, he initiated a joint project with the Department of Medical Physics at the Hebrew University—Leichter, Weinreb, and Hazan—and Prof. G. C. Robin. Their research focused on radiation-based quantitative methods for assessing bone health and demonstrated that gamma-ray Compton scattering is sensitive to bone mineral content and can detect early reductions in bone density.[111] However, these studies did not include a direct biomechanical evaluation of bone strength.

To address this gap, we introduced a biomechanical framework combining experimental testing with structural modelling. Fresh human femora were subjected to loading-to-failure, and the femoral neck was modelled as a composite beam consisting of a cortical shell and a porous cancellous core.[112]

Our findings established bone density as the most reliable predictor of femoral neck strength, outperforming bone mineral content and the Singh index.[113] Importantly, fracture strength showed strong dependence on cancellous bone density, underscoring the clinical relevance of trabecular structure in addition to cortical thickness.[112]

Porosity emerged as a critical factor in fracture initiation, linking microstructural deterioration to mechanical failure.[114] While bone density remains the best available indicator of mechanical competence, it is not sufficient alone: small reductions in density can result in disproportionately large strength losses, explaining fragility fractures in patients with seemingly adequate density.[115] These findings highlight that structural integrity is as important as mineral content in assessing fracture risk.

Subsequent work by Hayes and others on stress concentrations in osteoporotic bone, as well as later computational studies,[116] further supports the central role of porosity, trabecular architecture, and load distribution in determining bone strength and fracture risk.

Bone weakening due to metastases or osteoporosis (Harvard Medical School)

In 1989, I joined Professor Wilson ("Toby") C. Hayes at Beth Israel Hospital, Harvard Medical School, for a sabbatical focused on vertebral fragility. We examined how two distinct conditions—metastatic lesions and osteoporosis—affect vertebral strength using a common framework: linear elastic 3D Finite Element (FE) modelling with comparable geometry, material properties, and loading conditions. Outcomes included stress distributions, load sharing, and stress concentrations in cortical and trabecular bone.

For metastatic lesions, local defects were simulated within cancellous bone.[117] Results showed that even small lesions caused substantial stress redistribution, elevating cortical stresses without increased external load. Lesion location was critical: central lesions altered internal stress pathways, whereas eccentric lesions induced pronounced cor-

tical stress concentrations. These effects reflected disrupted load transfer rather than simple material loss.[117]
In osteoporosis, the problem shifts from local defects to global weakening.[116] With uniform material degradation, stress changes arise from systemic redistribution rather than localized perturbation. Together, these studies demonstrated that vertebral fragility is fundamentally a structural problem governed by altered load transfer, not merely reduced material strength.
Although early models represented trabecular architecture implicitly via elastic properties, they predicted that stiffness loss reroutes load and increases cortical stress. Subsequent micro-CT–based nonlinear FE models enabled explicit analysis of trabecular architecture and improved prediction of bone strength and failure.[118]
Overall, our early continuum FE work established that both focal and diffuse weakening drive vertebral fragility through changes in internal load transfer—a foundation later refined by microstructural modeling.

Biomechanical indeterminacy

From a mechanical perspective, the musculoskeletal system is inherently indeterminate, as the number of unknown muscle forces exceeds the available equations. In conditions of impairment, redundancy—and thus indeterminacy—decreases,[119] strengthening the relationship between higher and lower neural control levels. Examples include single- versus double-stance standing[120] and muscular fatigue.[43]
An extreme case occurs in complete lower-leg paralysis, where FES isolates a single active muscle. Under such conditions, external torque measurements allow direct resolution of muscle force, rendering the system determinate.[94,95,99]
Modelling the index finger incorporates multiple tendons with varying moment arms, leading to an indeterminate system (eight equations, fourteen unknowns).[121] This is solved by means of nonlinear optimization, typically minimizing muscle stress,[122] with squared stress enhancing tendon differentiation.[123] The model successfully reproduces realistic motion, tendon forces, and joint torques, and has been applied to estimate forces during violin fingering.[124]
To further address indeterminacy, Roth et al groups synergistic muscles in modelling ankle–knee dynamics during cycling.[88] Joint torques are obtained via inverse dynamics from kinematic and pedal-force data and distributed among muscles by means of optimization. Muscle-specific parameters—including bi-articular effects—are then identified from EMG–force relationships using first- and second-order transfer functions.

Sit-to-stand motion

In 1999, I spent a sabbatical year at the Hong Kong Polytechnic University (PolyU), working with Christina Hui-Chan in the Department of Rehabilitation Sciences. Among the research projects I pursued during this period, one particularly noteworthy topic was the problem of biomechanical indeterminacy, as encountered in movements such as sit-to-stand, both in healthy individuals and in people with Parkinson's disease.
Mechanical indeterminacy arises during the pre–seat-off phase of the sit-to-stand movement, when the lower limbs are simultaneously loaded by both foot–ground reaction forces and thigh–chair reaction forces. If the thigh–chair contact forces are unknown or neglected, inverse dynamics cannot uniquely and reliably determine the lower-limb joint torques, especially at the hip and knee prior to seat-off.
We addressed this indeterminacy by employing two force plates: one placed under the feet and another under the chair. Although the dynamic model yielded 15 scalar equations with up to 18 unknowns, the simultaneous measurement of both support reactions and their points of application effectively transformed the system from underdetermined to overdetermined. We then exploited this redundancy using an iterative least-squares procedure to obtain consistent solutions.[125,126]

Concluding perspective: integration across scales and systems

The studies presented in this paper span several decades and a wide range of topics, from uterine muscle mechanics to locomotion, rehabilitation, tissue engineering, and motor control. Despite this diversity, a number of unifying themes emerge that reflect both the evolution of the field and the continuity of the underlying questions.
A central conclusion is that biomechanical function cannot be understood in isolation at a single level. Whether in uterine contractions, joint loading, cartilage behavior, or locomotion, mechanical behavior reflects the interaction of structure, material properties, and active control. The integration of experimental measurement with mathematical modeling has therefore been essential throughout this work, allowing not only description but also interpretation of complex biological systems.
A second recurring theme is the importance of dynamic behavior. Systems traditionally regarded as static—such as standing posture or joint contact—are in fact dynamically regulated processes involving continuous feedback and adaptation. This insight has implications for both fundamental understanding and clinical practice, particularly in rehabilitation, where restoring function requires addressing not only structure but also control.
The concept of mechanical impedance provides a useful unifying framework. Rather than viewing stiffness, damping, and inertia as fixed properties, they emerge as task-dependent outcomes of neuromuscular activation. This perspective links diverse phenomena, including locomotion, perturbation response, fatigue, and manipulation, and highlights the role of the nervous system in shaping mechanical behavior to meet functional demands.
A related insight concerns biomechanical indeterminacy. The redundancy of the musculoskeletal system, often treated as a limitation, can instead be viewed as a source of adaptability and resilience. In healthy systems, redundancy enables flexible solutions to changing demands,

whereas in pathological conditions it is often reduced, leading to more constrained and less robust behavior. Understanding how biological systems exploit or compensate for this indeterminacy remains an important challenge.
Across applications—from gait analysis and prosthetic design to FES and tissue engineering—the work also underscores the importance of clinically meaningful simplification. While comprehensive measurement and modeling are inherently complex, practical advances often arise from identifying key variables that capture essential features of the system.
Looking forward, several directions appear particularly promising. These include the integration of multiscale models linking molecular and cellular processes to whole-body mechanics, the use of real-time sensing and feedback in rehabilitation technologies, and the continued development of personalized, patient-specific biomechanical analysis. Advances in imaging, computation, and data-driven methods will play an important role in these efforts.
In closing, biomechanics is inherently interdisciplinary, requiring the collaboration of engineers, clinicians, and scientists. The progress described here has depended on such collaborations, as well as on the contributions of students and colleagues over many years. The field continues to evolve, but its central aim remains unchanged: to understand how biological systems move, adapt, and fail, and to apply this understanding to improve human health and function.

List of abbreviations

ASYM, Asymmetry
BME, Biomedical Engineering
CG, Center of Gravity
CVA, Cerebro Vascular Accident
CCI, Cerebro Cranial Injury
CD, Current Density
IRC, Isometric Recruitment Curve
EMG, Electromyography
iEMG, Integrated Electromyography
GRF, Ground Reaction Force
MPF, Median Power Frequency
RSTV, Relative Sequence of the Tangential Vectors
SA, Sway Activity
WBI, Weight Bearing Imbalance

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Conflict of interest

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Ethics approval and consent to participate

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Availability of data and materials

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