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PHD THESIS

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CONTRIBUTIONS ON THE USE OF VIBRATIONS IN HYBRID ORTHOSES FOR THE OPTIMIZATION/REHABILITATION OF AFFECTED BIOMECHANICAL SYSTEMS

-Summary-

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INTRODUCTION

Biomechanical rehabilitation is a multidisciplinary field that brings together principles from engineering, medicine, and physiotherapy, aiming to restore the function of the musculoskeletal system. For patients affected by traumatic injuries, neurological disorders, or surgical interventions, orthoses play a central role, providing mechanical support and helping to regain mobility, strength, and the ability to perform daily activities [1].

Recent advances in medical technologies have led to the development of smart orthotic devices, capable of monitoring biomechanical responses in real time and actively assisting in the therapeutic process. Among these solutions, the integration of therapeutic vibrations into hybrid orthoses opens promising perspectives for personalized neuromuscular rehabilitation [2][23].

Controlled local vibrations can positively influence tissue regeneration, proprioception, and motor function, by transforming mechanical stimuli into biological responses at the cellular level [3]. Recent studies have shown their effectiveness in reducing spasticity and accelerating recovery, particularly in neurological conditions [4][6][7].

Conventional orthoses provide mechanical support, but the lack of active stimulation mechanisms limits their therapeutic impact, highlighting the need for modern solutions that integrate vibration-based approaches [13].

The present work addresses this need through a comprehensive approach that combines theoretical analysis, numerical modeling (FEM), and experimental testing of 3D-printed orthoses, aiming to identify natural vibration modes and provide a scientific basis for their use in biomechanical rehabilitation [14].

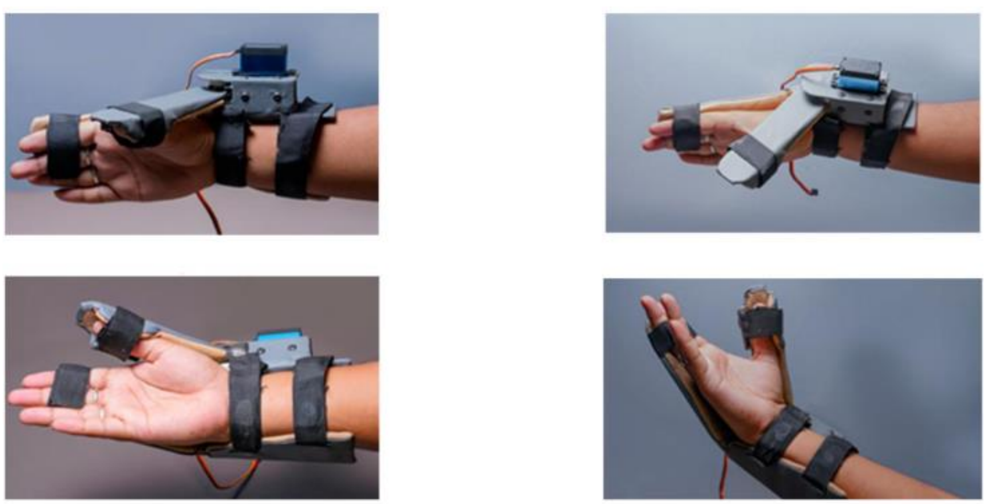

Figure 1. Portable orthotic device for hand rehabilitation, equipped with sensors and IoT capabilities for real-time monitoring and assessment

Research Objectives

The thesis aims to investigate the potential of using therapeutic vibrations in biomechanical rehabilitation through hybrid orthoses, with a focus on theoretical, numerical, and applied analysis aspects. In this regard, the specific objectives are:

1. To provide a literature review on the use of orthoses in musculoskeletal rehabilitation and the integration of vibrations as a therapeutic method.
2. To identify and classify vibration parameters (frequency, amplitude, duration, direction) and analyze their effects on biological tissues.
3. To perform numerical modeling of an orthotic system with vibration stimulation, using methods such as finite element analysis (FEM) or biomechanical simulation.
4. To simulate the biomechanical response of the human body to vibration application under controlled conditions, based on simplified hybrid orthosis models.
5. To validate the results obtained through simulation, by comparing with literature data or relevant experimental studies.

Applied Methodology

The methodological approach is complex, combining theoretical, numerical, and documentary aspects in order to provide a coherent evaluation of the impact of vibrations on the human body through a hybrid orthosis. The methodological structure can be summarized as follows:

1. Review of the specialized literature, with emphasis on the biomechanical applications of therapeutic vibrations, the typologies of orthoses, and the evolution of hybrid orthotic devices.
2. Identification and classification of vibration parameters used for therapeutic purposes, according to frequency, amplitude, duration, and direction.
3. Numerical modeling of a virtual orthotic system that integrates vibration functions, using methods such as finite element analysis (FEM) and biomechanical simulations;
4. Simulation of the dynamic behavior of the model under vibration exposure and analysis of force distribution on musculoskeletal structures in controlled conditions.
5. Comparison of the obtained results with data from the specialized literature, in order to validate the consistency of the model and highlight the potential benefits or limitations of this technology.

CHAPTER 1: THEORETICAL FOUNDATIONS OF BIOMECHANICS AND ORTHOTICS

Orthotics is a fundamental field of biomechanical rehabilitation, with the role of supporting, aligning, or correcting the function of affected body segments through external devices called orthoses. These devices make a significant contribution to restoring motor function in patients with various traumatic, neurological, degenerative, or post-surgical conditions. For their effective application, a deep understanding of the biomechanics of the musculoskeletal system is required, since each orthosis acts directly on the body, influencing force transmission and movement control [13][15][18].

The human musculoskeletal system is composed of bones, joints, muscles, tendons, and ligaments, which work together to ensure mobility and maintain posture. An orthosis interacting with this system must respect functional alignments and transmit forces in a biomechanically efficient way. For example, aligning the orthosis with the joint rotation axes prevents unnecessary strain on soft tissues and ensures patient comfort [15][18][23].

The design of orthotic devices is based on several biomechanical principles, among which the principle of the lever is particularly important. In this context, the orthosis functions as a first-, second-, or third-class mechanical lever, depending on the positioning of the fulcrum, resistance, and force. Such levers are integrated, for instance, in knee orthoses to reduce muscle effort, or in spinal orthoses, where support forces are redistributed to areas more tolerant to pressure [25][26].

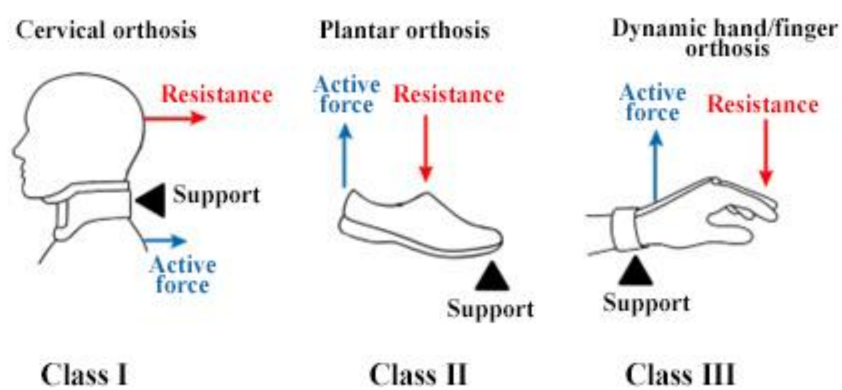

Figure 2. Representative examples of the three types of mechanical levers applied in orthotic devices [25][26]

Another frequently applied principle is the three-point pressure system, through which the orthosis exerts forces in a controlled manner to correct joint or postural deviations. The simultaneous application of two forces on either side of the segment, along with an opposing counterforce, generates a corrective biomechanical moment. This approach is found, for example, in thoracic orthoses for scoliosis, knee orthoses for varus or valgus deviations, and corrective devices used in pediatric rehabilitation [18][23][25].

An essential aspect for the functionality of an orthosis is the correct distribution of pressure at the interface between the device and the skin. Excessive pressure concentrated on a small area can lead to irritation, ulcers, or even tissue damage. To prevent these effects, the contact surface of the orthosis is designed to distribute mechanical pressure evenly. This is achieved by adapting the internal shape to the patient's morphology, using deformable materials, and analyzing force distribution through digital simulation [14][39].

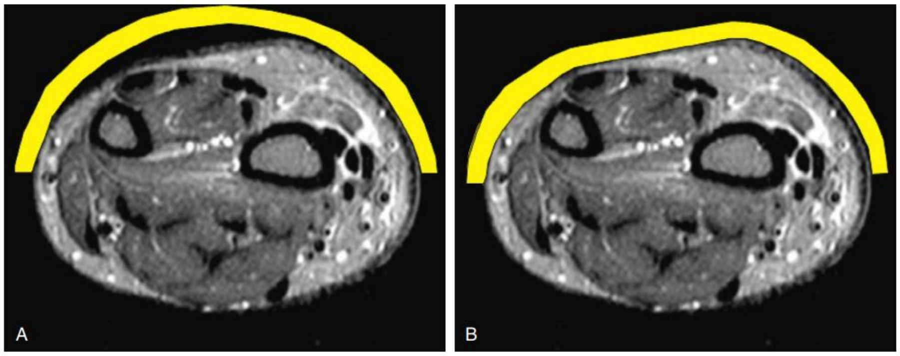

Figure 3. Illustration of how the shape of the orthosis influences the distribution of contact pressure on the forearm

In recent decades, technological progress has enabled the integration of advanced materials into orthoses: lightweight composites, shape-memory materials, gel inserts, or smart textiles. The choice of material influences not only the weight and comfort of the orthosis, but also its ability to transmit or absorb forces applied during walking, support, or functional re-education [33][38].

From the perspective of functional adaptability, orthoses can be classified according to their therapeutic purpose: immobilization, stabilization, functional, corrective, or compensatory. In addition, depending on the degree of mobility allowed, they can be static (immobile), semi-dynamic (with limited mobility), or dynamic (with controlled freedom). This classification influences both the type of conditions for which they are recommended and the way the patient interacts with the device [18][23].

The thesis reviews the main conditions that require orthotic treatment, grouped by their causes: trauma (fractures, sprains, ligament injuries), neurological disorders (stroke, cerebral palsy), degenerative diseases (osteoarthritis, spondylosis), congenital and pediatric conditions (hip dysplasia, clubfoot), post-surgical indications (postoperative immobilization), and other special clinical cases.

For each category, examples of orthoses are presented: the AFO (Ankle-Foot Orthosis) for neurological conditions of the lower limb, the Pavlik harness for hip dysplasia, Denis Browne splints for foot position correction, and lumbosacral orthoses used in post-surgical recovery. These are designed to meet precise biomechanical requirements, adapted to each clinical context.

Another important aspect discussed in this chapter is the role of numerical modeling in orthosis design. Using methods such as finite element analysis (FEM), the structural behavior of orthoses can be simulated under real usage conditions. This approach makes it possible to optimize the shape, evaluate stress distribution, prevent mechanical failure points, and achieve full customization of the orthotic device [20][21][39].

The integration of these principles and modern technologies into contemporary orthotics has led to increasingly efficient, adaptable, and comfortable devices. These not only help maintain joint function and prevent further damage, but also actively support neuromuscular recovery, especially when combined with active control systems or stimulation technologies, as in the case of hybrid orthoses analyzed in the following chapters of this thesis.

CHAPTER 2: THE INFLUENCE OF VIBRATIONS ON THE HUMAN BODY

Vibrations are an omnipresent physical phenomenon, defined as the mechanical oscillations of a system around an equilibrium position [13]. In a biomechanical context, they can directly influence human health, having both beneficial and harmful effects depending on the exposure parameters and the individual characteristics of the body [14][15]. The main parameters used to characterize vibrations are frequency, amplitude, and acceleration. The relationship between displacement, velocity, and acceleration is expressed by the formula:

$$a(t) = -\omega^2 \cdot x(t), \qquad \omega = 2\pi \tag{1}$$

where $x(t)$ is the displacement, and ω is the angular velocity.

Exposure of soft tissues to vibrations leads to structural and functional changes. The effects include increased muscle tension, circulatory changes, and influences on tactile sensitivity [17][18]. At the bone level, the response depends on frequency and amplitude, with the potential to stimulate osteogenesis at controlled values, but also with the risk of microtrauma and degeneration when exposure is excessive [11].

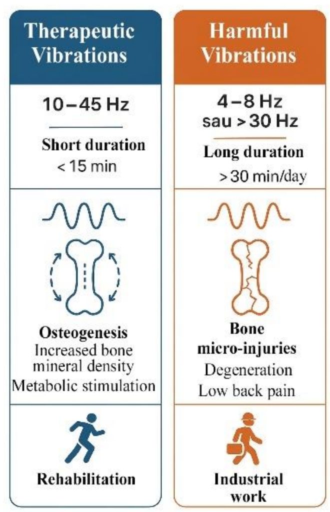

Figure 4. Bone tissue response to mechanical vibrations depending on frequency and duration

Each biomechanical structure has a natural frequency of oscillation. This can be calculated (in simple models) using the formula:

$$\omega_0 = \sqrt{\frac{k}{m}} \tag{2}$$

where k is the stiffness of the segment [N/m], m is the equivalent mass [kg] si ω_0 is the natural angular frequency [Hz].

The human body exhibits resonance frequencies at different segments, which leads to variability in the effects. The lumbar region is sensitive to vibrations of 4–6 Hz, while the head and neck respond mainly to 20–30 Hz [18]. These values are essential for understanding how vibrations can cause selective discomfort or injury.

The soft tissues of the body, such as muscles, tendons, ligaments, and cartilage, respond differently to vibrations depending on the physical parameters and the individual characteristics of the subject [8]. At low frequencies (5–20 Hz), muscle reflexes and slow fibers are stimulated, which are useful for maintaining posture, while higher frequencies (30–100 Hz) activate fast fibers, enhancing strength but also increasing the risk of fatigue. In well-calibrated therapeutic regimes (20–50 Hz, small amplitudes, short exposures), vibrations can stimulate circulation, reduce excessive tone, and support muscle recovery [9].

On the other hand, uncontrolled or prolonged exposures can be harmful. Tendons and ligaments, although partially absorbing vibration energy, may suffer micro-injuries or local inflammation, especially at frequencies above 60 Hz [3][8][11]. Articular cartilage, with its limited regenerative capacity, is vulnerable to structural degradation; however, low and well-dosed vibrations can stimulate chondrocyte regeneration [4][18]. Overall, the beneficial or harmful effects depend on the balance between frequency, amplitude, and duration, which makes the adjustment of vibration parameters essential for the safety and effectiveness of clinical interventions [6][8][11].

Parameter	*Predominant Tissue Response*
< 20 Hz	neuromuscular reflexes, postural activation
30-60 Hz	peak EMG activation, metabolic stimulation
>60 Hz	risk of fatigue, local inflammation, microtrauma
< 10 min	predominantly positive effects (if parameters are optimal)
> 30min	onset of adverse reactions and local overstrain

Table 1. The response of soft tissues is significantly influenced by vibration characteristics.

These tissue mechanisms also explain how clinical syndromes develop in individuals professionally exposed to vibrations. Whole-body vibration (WBV) is commonly encountered among heavy vehicle operators and may lead to low back pain, intervertebral disc damage, and chronic muscular disorders. In contrast, local exposure (HAV), typical of using electric or pneumatic tools, manifests as hand-arm vibration syndrome (white finger), peripheral neuropathies, and reduced fine manipulation ability [26].

These clinical manifestations have led to the need for clear health protection regulations for exposed workers. In this regard, international bodies such as ISO and HSE have established strict standards that define permissible thresholds and evaluation methods for both whole-body vibration (WBV) and hand-arm vibration (HAV) [11][44]. ISO 2631-1:2001 sets action values and limits for daily exposure to whole-body vibrations, while ISO 5349-1:2001 establishes similar thresholds for local exposure, associated with the risk of developing HAVS [11][44]. Compliance with these limits, complemented by practical strategies such as the use of vibration-damping materials, ergonomic suspension systems, and rational work scheduling, forms the basis of preventing occupational complications [44][50].

Besides risks, vibrations can also be used for therapeutic purposes when applied in a controlled manner. Low-frequency, low-amplitude vibrations have proven effective in stimulating muscle contractions, increasing bone density, and improving balance. However, improper application or excessive exposure may cause adverse effects such as micro-injuries, inflammation, or joint degradation.

An emerging field is the integration of vibrations with Functional Electrical Stimulation (FES). This method involves inducing muscle contractions through electrical impulses applied to muscles or nerves. Combining FES with vibrations allows the integration of mechanical and electrical stimuli, which can enhance the effectiveness of neuromuscular rehabilitation.

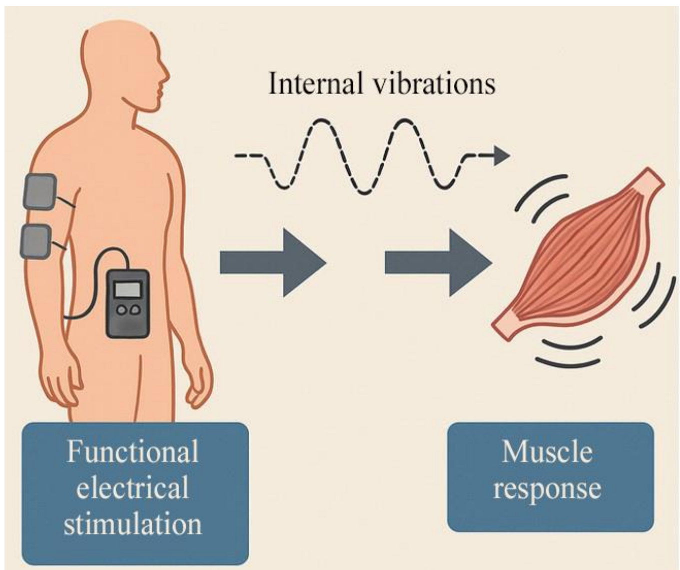

Figure 5. Induction of therapeutic vibrations through electrical stimulation

Overall, vibrations have a dual nature: they can be sources of severe occupational disorders, but also valuable tools in rehabilitation when application parameters are carefully controlled. Understanding the underlying mechanisms, complying with standards, and integrating modern technologies are essential prerequisites for harnessing the therapeutic potential of vibrations and minimizing associated risks, aspects detailed within this chapter.

CHAPTER 3: THE USE OF ORTHOSES IN BIOMECHANICAL REHABILITATION

This chapter provides a detailed analysis of how orthoses are used as tools for support, correction, and movement facilitation in biomechanical rehabilitation therapies. In the modern context, an orthosis is no longer just a passive support, but an active component integrated into multidisciplinary treatments that include electrostimulation, functional exercises, and sensory feedback [13].

Depending on the objective, orthoses may serve to stabilize, correct, or provide functional support: in acute injuries such as fractures or surgical interventions, they protect healing structures, while in neurological pathologies, such as cerebral palsy or stroke, they help maintain proper biomechanical alignment and prevent secondary deformities [15]. At the same time, modern orthoses actively contribute to neuromotor re-education by supporting correct movements, reactivating residual musculature, and improving postural stability and gait; for elderly patients or individuals with disabilities, they reduce energy expenditure and the risk of falls [18][23][24].

The prescription of orthoses has become part of a multidisciplinary approach in which orthopedic surgeons, rehabilitation physicians, physiotherapists, and biomedical engineers collaborate to personalize the device according to therapeutic goals and the patient's specific characteristics [23]. In this context, recent technological advances have radically changed the design and manufacturing of orthoses. Whereas in the past they were standardized devices, today they are fully customizable thanks to digital technologies, computer-aided modeling, and the integration of numerical methods. Finite element analysis (FEM) enables simulation of biomechanical behavior, highlighting stress zones, stiffness, and functional efficiency during walking or segmental movements, thus reducing the risk of excessive pressure points and instabilities [20][21].

A key innovation is the adoption of CAD/CAM technology, which allows three-dimensional scanning of the body segment and simulation of pressure and force distribution before prototype fabrication [14][20]. In parallel, 3D printing has become a widely used solution in clinical orthotics, enabling the creation of lightweight, fully customized orthoses with differentiated stiffness zones, which are difficult to achieve through traditional methods [18][58]. The stages of this process are illustrated in Figure 6, which shows plantar scanning, pressure analysis, and 3D printing of customized orthoses [57].

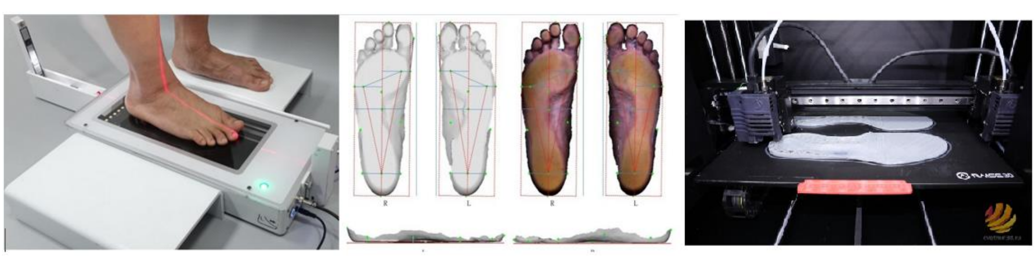

Figure 6. Plantar scanning, pressure analysis, and 3D printing: stages in the fabrication of customized plantar orthoses [57]

The materials used have also undergone significant diversification. From metallic alloys and rigid plastics, the field has moved toward technical polymers, carbon fibers, composite materials, and shape-memory foams, which combine structural rigidity with flexibility and biocompatibility [24][38]. These innovations not only increase comfort but also enable the integration of sensors, transforming orthosis design into a predictive, individualized process with clear benefits for therapeutic efficiency and broader clinical applicability.

These advances in design and materials have paved the way for the development of smart orthoses. The miniaturization of sensors and the increase in processing power have made it possible to integrate modules capable of measuring pressure, acceleration, joint position, or muscle activity, providing real-time feedback and allowing therapy to be adapted to the patient's needs [14][18][38]. Among the most relevant applications is the integration of Functional Electrical Stimulation (FES), which can be automatically activated to assist gait or stabilize joints.

Functional Electrical Stimulation (FES) is already a well-established method for activating paralyzed or weakened muscles by applying electrical impulses to peripheral nerves. When integrated into orthoses, this technique has led to the development of hybrid devices that combine mechanical support with active motion control. Through this approach, the patient benefits not only from support but also from direct facilitation of walking and other complex functional activities [18][38].

Relevant clinical examples include the Bioness L300 Go and WalkAide®, devices capable of detecting gait phases through sensors and applying electrical stimulation to the dorsiflexor muscles, thereby correcting foot drop. Published results show clear positive effects: increased walking speed, reduced energy cost, and greater independence for patients with neurological conditions, including post-stroke or multiple sclerosis [61].

Figure 7. The Bioness L300 Go device used for functional electrical stimulation of the dorsiflexor muscles during walking [61]

Moreover, the integration of FES into hybrid orthoses represents a significant advancement compared to traditional passive devices. These solutions not only provide mechanical assistance for movement but also actively stimulate the muscles, promoting coordination and neural reorganization. By including them in modern rehabilitation protocols, patients increase their independence and reduce reliance on external assistance, confirming both the clinical effectiveness and the potential of these technologies to become a therapeutic standard in neuromotor disorders [61][62].

Building on the integration of electrical stimulation and sensory feedback, hybrid orthoses have naturally evolved into partial exoskeletons and robotic devices, enabling active and real-time adaptive rehabilitation. Unlike classic EMG orthoses, which depend on residual muscle activity, these systems use electric or pneumatic actuators and a wide range of sensors (EMG, inertial, positional), making them suitable for supporting both patients with minimal activation and those with severe paralysis. Clinical studies confirm their benefits: from reducing spasticity and improving arm movements through the integration of EMG and inertial sensors, to the use of neural networks or even EEG signals for recognizing movement intentions and adjusting the level of assistance [65].

A notable advancement is the emergence of flexible and biocompatible sensors, capable of providing precise biomechanical data without restricting mobility, thus optimizing actuator control and patient comfort [72]. However, the performance of these devices depends on critical technological factors such as sensor accuracy, actuator type, response latency, weight, and portability. Electric actuators ensure precise control but can be noisy, pneumatic ones offer more natural movements but are difficult to miniaturize, while passive systems remain limited to static support [72]. Overall, these advanced systems combine real-time personalization with the stimulation of neuroplasticity, making them promising tools in modern functional rehabilitation.

Figure 8. Upper-limb exoskeleton equipped with deformable sensors, designed to adjust the level of assistance according to the patient's movements and effort [72].

Chapter 3 highlighted the evolution of orthoses from simple passive devices to intelligent and robotic systems capable of integrating sensors, electrical stimulation, and real-time feedback. Their effectiveness depends on a balance between biomechanical design, technology, and human and clinical factors. Essentially, the role of these devices is no longer limited to providing support but to becoming active tools for recovery and independence, confirming the importance of a personalized and multidisciplinary approach in modern rehabilitation [75].

CHAPTER 4: COMPUTER-AIDED ANALYSIS OF VIBRATIONS IN ORTHOTIC SYSTEMS

Chapter 4 examines the ways in which vibrations applied in orthoses can be studied and optimized using numerical simulations. These methods allow for a detailed understanding of mechanical phenomena and the identification of optimal design solutions without the need for repeated and costly experiments [79].

One of the fundamental tools is the Finite Element Method (FEM), widely used to simulate the biomechanical behavior of orthotic structures. By discretizing a segment or joint into finite elements, it becomes possible to highlight areas of mechanical stress, stiffness, and vibration distribution. FEM makes it possible to anticipate how the materials and geometry of the orthosis influence the transmission of vibrations to the body, thereby reducing the risk of excessive pressure points or instabilities [20][21].

An illustrative example is the modeling of the wrist joint using an FEM mesh, which enabled the analysis of stress distribution and the optimization of orthotic design for a more precise user-specific adaptation [81].

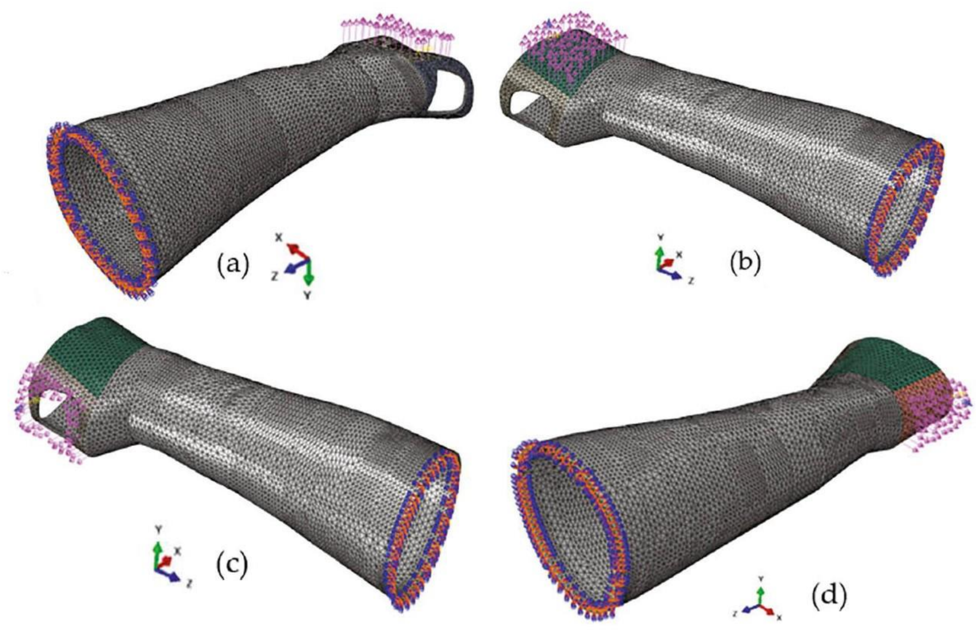

Figure 9. Finite element mesh, boundary conditions, and load application directions corresponding to the four main wrist joint movements: (a) flexion, (b) extension, (c) radial deviation, (d) ulnar deviation [81].

In addition to FEM, the Boundary Element Method (BEM) provides advantages in analyzing structure–environment interactions, particularly in situations where vibration transmission involves both tissues and the external environment. This method reduces model complexity by discretizing only the surface, making it suitable for acoustic simulations and for evaluating vibrations in areas that are difficult to model with FEM [82][85]. Recent literature examples show that BEM, when combined with FEM, leads to a more realistic assessment of vibrational phenomena affecting both the body and the orthotic device.

Another approach is the Discrete Element Method (DEM), used to simulate the behavior of structures composed of particles or segments with multiple contacts. In biomechanical orthotics, DEM has been applied to understand the contact between the orthosis surface and the skin, as well as how vibrations are dissipated through materials with variable stiffness. This method is particularly useful for designing customized orthoses that include adaptive foams or composite materials [86].

Overall, these numerical methods provide complementary perspectives, each with its own advantages depending on the type of analysis required. FEM is more accurate for assessing structural stress, BEM for boundary interactions, and DEM for complex and dynamic contacts. Together, they create a detailed picture of how vibrations are transmitted and controlled in orthoses, forming the basis for further design optimization.

In addition to numerical methods, this chapter also highlights the importance of modal analysis, used to determine the natural frequencies and vibration modes of orthotic structures. Identifying these frequencies is essential to avoid resonance, a phenomenon that could amplify vibrations transmitted to the body. By correlating modal data with FEM simulations, designers can adapt geometry and materials to achieve a safe and efficient biomechanical response [82][84].

Building on these theoretical and experimental foundations, the classical optimization of vibrations in orthoses relies on deterministic methods, relatively simple yet accurate, aimed at either maximizing therapeutic effect or reducing discomfort. Parametric analysis gradually modifies variables such as frequency, amplitude, or vibration source location, evaluating their impact on mechanical behavior and patient comfort. For example, the choice of where vibration is applied on the orthosis surface can directly influence the efficiency of transmission to target tissues, while frequency adjustment helps avoid resonance, a potentially harmful phenomenon [80][89].

A key aspect is the adjustment of vibration amplitude, which must remain within a range that is both safe and therapeutically effective. Figure 10 illustrates this relationship, distinguishing between the sub-therapeutic zone (low efficiency), the optimal zone (effective and safe), and the unsafe zone, where excessive amplitudes may cause discomfort or adverse effects [10][23][42]. Through such approaches, even without complex technologies or advanced algorithms, the performance of an orthosis can be significantly improved from the early stages of design.

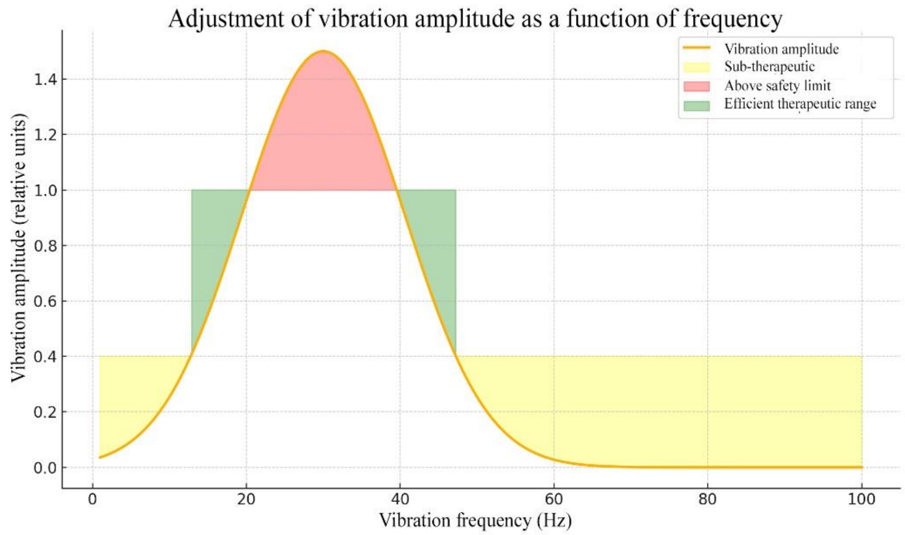

Figure 10. Representation of the relationship between vibration frequency and transmitted amplitude, highlighting the efficient therapeutic range and safety limits

Beyond parametric optimization, the chapter also introduces topological optimization, an advanced computer-aided design method. This approach is not limited to modifying predefined parameters or shapes but allows the redistribution of material within a structure, generating completely new configurations. In the orthotic field, this method is extremely valuable, as orthoses must simultaneously meet multiple requirements: efficiently transmit vibrations, remain as lightweight as possible, ensure user comfort, and adapt to the patient's morphology [79][81].

By applying topological optimization, areas of an orthosis can be identified where material may be removed without compromising structural integrity, thus reducing the device's weight and improving portability. At the same time, this method helps direct vibrations toward therapeutically beneficial regions while preventing their propagation to sensitive or painful areas [23][59]. Figure 11 illustrates how the outcome of a topological optimization process, initially obtained as a discretized model, can be transformed into a complete CAD geometry and later into a functional orthosis manufactured from modern materials such as carbon fiber–reinforced PETG [91].

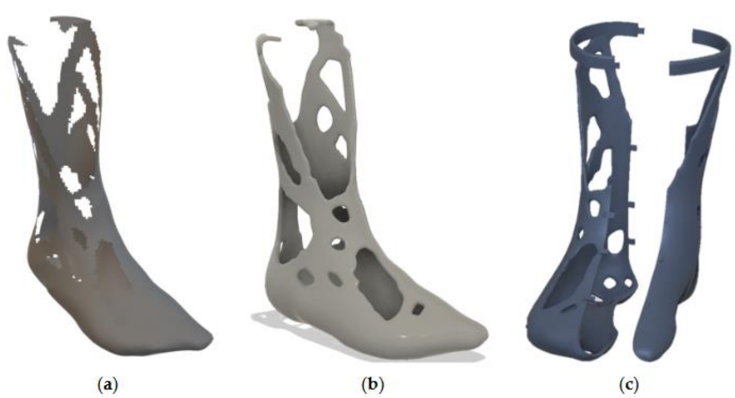

Figure 11. Stages of transforming a topologically optimized model into a wearable ankle orthosis

Combined with modal analysis and finite element simulations, topological optimization becomes an integrated design tool, capable of providing both the necessary mechanical stability and a vibrational response adapted to therapeutic objectives. Thus, this technique opens new perspectives for the development of personalized, efficient, and innovative orthoses that go beyond the limitations of traditional solutions [80][81].

A crucial step after simulation and optimization is experimental validation. No matter how advanced, numerical models prove their usefulness only when the obtained data are confirmed through real biomechanical measurements, carried out either on patients or on experimental models. In the reviewed studies, this validation was performed using accelerometers mounted on a 3D-printed orthosis, fixed on a vibrating platform, and the obtained results were compared with those calculated analytically, confirming the accuracy of the applied models [13][15].

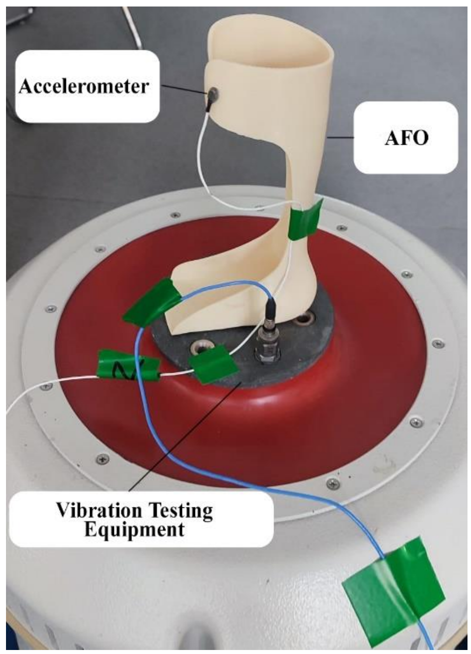

Figure 12. Experimental setup for testing vibrations transmitted through a 3D-printed AFO orthosis

Overall, Chapter 4 highlights the central role of numerical simulations in the design and evaluation of hybrid orthoses with integrated vibrations. FEM, BEM, DEM, and modal analysis provide a detailed picture of the interaction between vibrations, materials, and the human body, while optimization combined with experimental validation ensures clinical applicability. This synthesis between digital tools and practical verification forms the foundation for developing a new generation of orthoses capable of maximizing therapeutic effects and reducing the risks of vibrational exposure.

CHAPTER 5: INTEGRATION OF VIBRATIONS INTO HYBRID ORTHOSES – DESIGN AND OPTIMIZATION

Chapter 5 introduces the concept of the hybrid orthosis, defined as a device that combines the passive mechanical structure of traditional orthoses (shell, joints, fastening elements) with active or intelligent systems, such as vibration sources, sensors, or actuators. This approach marks the transition from simple mechanical support of the body segment to active interaction with the patient, including targeted stimulation, data collection, and real-time adaptation [2][18].

An illustrative example is shown in Figure 13a, which depicts the NuroSleeve orthosis, a customized device that integrates a 3D-printed structure, a central controller, multiple activation options (joystick, EMG, voice, inertial sensors), and a functional electrical stimulation (FES) module [92]. By comparison, Figure 13b shows a classic passive orthosis, which provides only mechanical support of the joints, without interaction or adaptation [93]. This difference highlights the transition from passive support to actively assisted rehabilitation, characteristic of hybrid orthoses

Figure 13. Comparison between two types of orthoses: (a) customized hybrid orthosis NuroSleeve, with active control and integrated stimulation [92]; (b) classic passive wrist orthosis [93].

Compared to robotic exoskeletons, which can generate movement independently of the user and have a high degree of autonomy, hybrid orthoses are not designed to replace muscular effort but rather to assist and stimulate existing functions. They are mainly used in rehabilitation to support the resumption of movement, muscle activation, and the correction of postural imbalances. Clinical studies show that integrating active stimulation with passive support yields promising results in post-stroke treatments, neurological conditions, and the recovery of locomotor function [18][23].

A central role in transforming a conventional orthosis into a hybrid device is played by active rehabilitation systems. Among the most widely used technologies are:

- Functional Electrical Stimulation (FES), which triggers coordinated muscle contractions and is successfully used for gait retraining or hand movements;
- EMG sensors, which capture muscle signals and allow real-time adaptation of stimulation levels;
- as well as alternative control methods – inertial sensors, joysticks, or voice commands – useful when the muscle signal is not available.

Such solutions personalize rehabilitation and facilitate the patient's active participation, transforming the orthosis into an intelligent and adaptable tool.

A key element in the development of hybrid orthoses is the integration of vibrations as a therapeutic method. The chapter emphasizes that vibrations can be used both for stimulating muscles and proprioception, as well as for pain relief and increased joint mobility [4][10]. In orthoses, this integration involves selecting miniaturized vibration sources (motors, piezoelectric actuators) and strategically positioning them to transmit the stimulus to the target areas without causing discomfort [18][38].

The basic principle lies in the selective transmission of vibrations. Thus, the geometry of the orthosis and the materials used are adjusted to direct the mechanical stimulus toward the desired region, avoiding propagation to sensitive structures. For example, in ankle orthoses, vibrations can be focused on the dorsiflexor muscles to prevent foot drop, while protecting the adjacent joints.

Moreover, the integration of vibrations in orthoses is not limited to passive stimulation. By combining them with sensory feedback and control algorithms, the devices can adapt the intensity and location of the stimulus in real time, depending on the patient's response. This bidirectional interaction transforms the orthosis from a simple support into an intelligent rehabilitation system, capable of personalizing therapy and maximizing recovery efficiency..

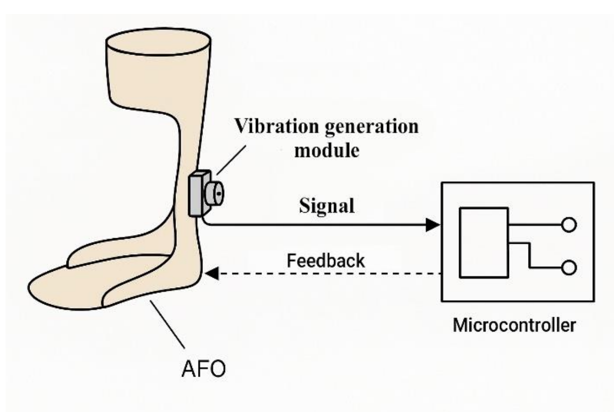

Figure 14. Conceptual diagram of a hybrid orthotic system with controlled vibration generation

The transmission of vibrations largely depends on the materials used. An ideal material should combine the stiffness required for efficient propagation of the mechanical wave with a moderate damping capacity, so that the stimulus is transmitted to the tissues without causing discomfort [13][18]. Figure 15 compares the damping coefficients of the main materials used in orthosis construction, illustrating the differences between rigid, flexible, and textile ones.

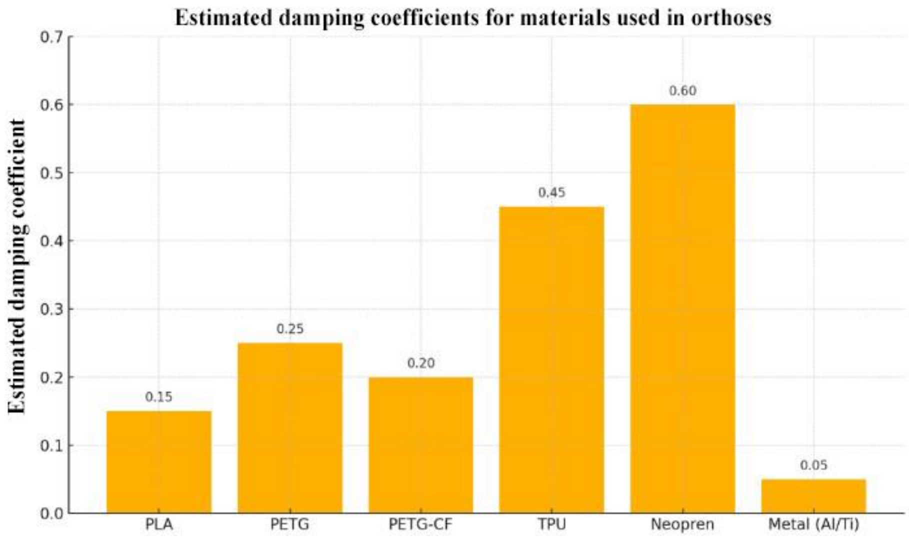

Figure 15. Estimated damping coefficients for the main materials used in orthosis construction

In this work, the focus was placed on two fundamental materials in additive manufacturing of orthoses: PLA, known for its rigidity but fragile under dynamic conditions, and PETG, which provides a compromise between strength and flexibility. Their selection enables a direct evaluation of how damping coefficient and stiffness influence the transmission of vibrations to anatomical segments [26][81][91].

Beyond intrinsic properties, the geometry of the orthosis and the configuration of 3D printing play an important role: wall thickness, layer orientation, or infill density may facilitate the propagation of the mechanical wave or, on the contrary, introduce attenuation effects. The fastening method on the limb (bands, straps, or rigid systems) determines the degree of contact with the skin and the level of vibration perceived by the patient [81][91].

In conclusion, the choice of material and constructive configuration targets not only mechanical strength and wearing comfort but also the efficiency of therapeutic vibration transmission. Optimizing these factors is essential for obtaining modern orthoses adapted to the functional objective—whether active stimulation, protection, or the integration of smart modules. These aspects lay the groundwork for the detailed analysis in Chapter 6, where the vibration behavior of 3D-printed AFO prototypes made of PLA and PETG will be investigated through FEM simulations and experimental tests [13][26][91].

CHAPTER 6: PRELIMINARY EXPERIMENTAL RESEARCH ON THE INTEGRATION OF VIBRATIONS INTO ORTHOTIC DEVICES FOR REHABILITATION

Chapter 6 describes the first experimental research dedicated to evaluating how mechanical vibrations can be integrated into 3D-printed orthoses. The objective was to analyze the vibration behavior of prototypes made from two materials frequently used in additive manufacturing, PLA and PETG, in order to highlight the influence of both material and geometry on the transmission of mechanical stimuli.

Two types of samples were tested: AFO (Ankle–Foot Orthosis) devices and simple parallelepiped-shaped specimens, both entirely manufactured from PLA and PETG. This dual approach allowed the observation of both the intrinsic vibrational properties of the materials and the effect that the complex shape of an orthosis has on vibration propagation.

In parallel with the experimental tests, numerical simulations were developed based on the actual geometry of the orthoses, using platforms dedicated to mechanical analysis. These virtual models were designed to reproduce the physical testing conditions, serving to validate the obtained results and to identify behaviors that are more difficult to capture experimentally.

Additionally, to complement the analysis, a practical visit was conducted to a company specialized in customized 3D-printed orthoses. The experience provided valuable insights into the manufacturing stages, from patient scanning to the final product, as well as the challenges encountered in clinical application.

By combining experimental tests with numerical simulations and observations from professional practice, the research creates a solid framework for evaluating how materials and orthosis geometry can influence the transmission of vibrations and their therapeutic relevance.

The numerical analysis involved the use of two specialized software platforms – FEBio, focused on biomechanical applications, and SolidWorks Simulation, a comprehensive commercial platform with extensive engineering applications.

In FEBio, the 3D models of the AFO orthoses and simple specimens were imported, defined with the mechanical properties of PLA and PETG, and discretized into finite elements. The fixation conditions were established to reproduce the experimental setup, and the dynamic behavior was analyzed through mechanical impulses and the monitoring of accelerations at key points.

The simulation made it possible to highlight the natural vibration modes and the general trends, even though some differences compared to the physical tests were inevitable.

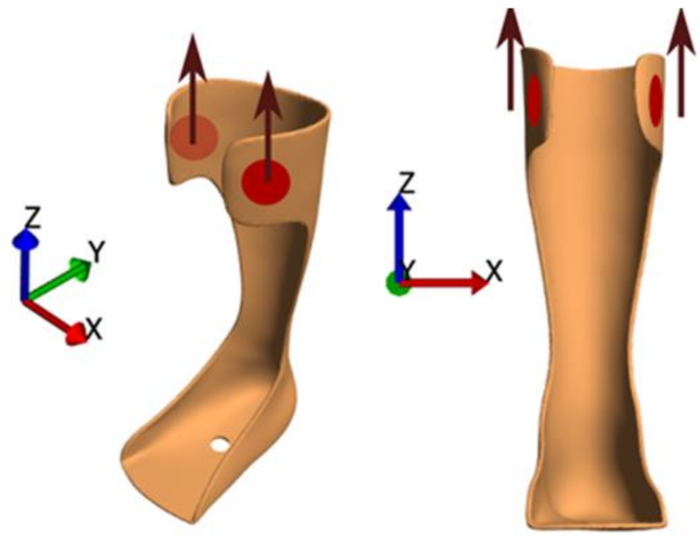

Figure 16. Configuration of mechanical impulse application in the vibration analysis performed with FEBio

The simulation with FEBio required some additional adjustments, since in harmonic analysis it was essential that the imposed acceleration remain constant regardless of frequency. To achieve this, the mathematical relationships between displacement and acceleration were manually introduced, with displacement amplitude decreasing as frequency increased. In this way, the excitation could be applied consistently within the 15–115 Hz range, and the results obtained were comparable and relevant to the vibrational behavior of PLA and PETG orthoses. The analysis showed good agreement between the mechanical impulse method and the harmonic method, with the identified resonance regions being similar. This coherence confirms the robustness of the numerical model and provides confidence in the reliability of the simulations for characterizing the vibrational behavior of orthoses.

In SolidWorks Simulation, the analysis followed a similar approach, but with more versatile preprocessing and visualization tools, which facilitated comparisons between the materials used and the geometry of the parts. The platform automatically manages vibration laws, so no manual interventions were required as in FEBio. A Frequency analysis was used, through which the natural vibration modes were identified, determined exclusively by the mass and stiffness of the structure.

The results showed that PETG AFO orthoses exhibited lower natural frequencies (22–107 Hz), reflecting moderate stiffness, while PLA, a stiffer material, generated higher values (26–124 Hz). The same trend was confirmed in the simple specimens, where PLA consistently recorded higher natural frequencies than PETG.

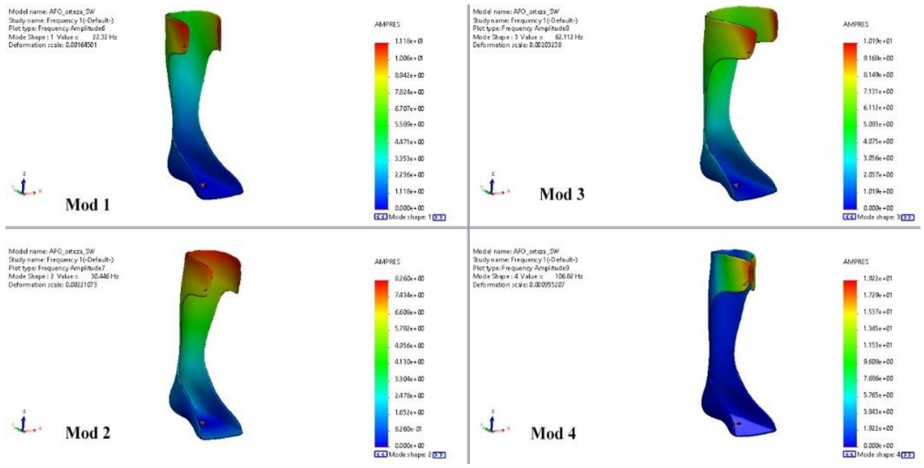

Figure 17. Modal shapes of the PETG orthosis (modes 1–4)

Thus, the combination of the two analysis platforms—FEBio, with its biomechanical rigor, and SolidWorks, with its engineering-oriented accuracy—provided a complementary perspective on the vibrational behavior of orthoses. The choice of modal analysis in SolidWorks proved useful for clearly identifying natural modes and for validating the trends observed both in FEBio simulations and in experimental tests.

To verify the actual behavior of the 3D-printed AFO orthoses and specimens, a series of experimental tests was carried out at the Research and Experimentation Center for Acoustics and Vibrations (Măgurele).

The tests consisted of applying controlled vibrations in the range of 0–2000 Hz using a fully calibrated system (Tiravib shaker, DYTRAN accelerometer, and Dewesoft acquisition), while the response of the samples was analyzed using the Fast Fourier Transform (FFT)

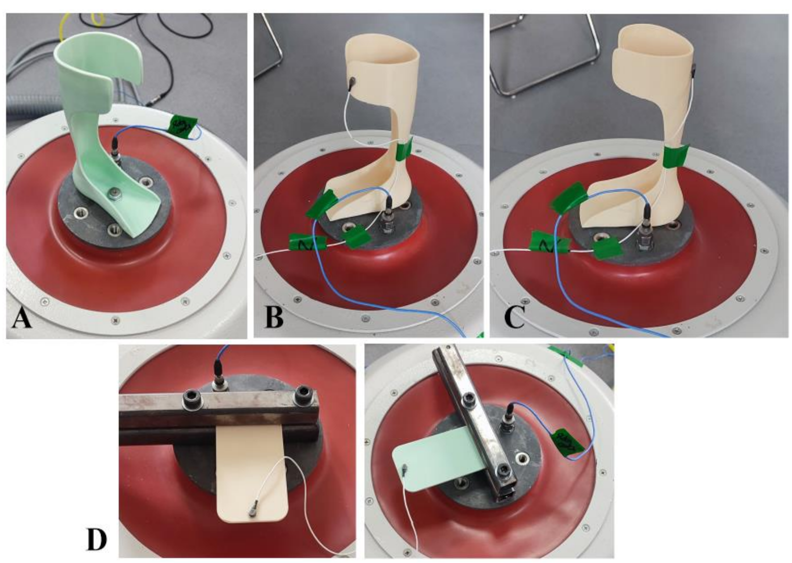

Figure 18. Positioning of the samples on the shaker and of the accelerometer during testing

Orthoses printed entirely from PLA and PETG, as well as parallel test specimens made from the same materials, were tested. The accelerometer was mounted successively on the lateral and posterior sides to capture differences in response depending on positioning.

The results confirmed the trends highlighted in the numerical simulations: the first natural mode of the PLA orthoses was identified around 35 Hz, close to FEM predictions, while for PETG the values were lower (24–26 Hz), reflecting the lower rigidity of the material. The differences observed between simulations and reality can be explained by the layered structure and infill percentage of the 3D-printed parts, but the overall consistency between the methods validates both the numerical model and the relevance of the experimental tests.

To complement the theoretical and experimental sections, the work also included a practical documentation stage carried out in collaboration with the company EITguilde SRL and the technical partner H-Shift Engineering, specialized in the design and production of customized 3D-printed orthoses. Discussions with Eng. Mihail Hornea provided a realistic perspective on the challenges of engineering practice and on the technological workflow used in orthosis manufacturing.

The following key aspects were highlighted:

- Material selection (PLA Tough, PETG, PETG-CF, TPU), depending on rigidity, pressure zones, and patient comfort.
- Design process: 3D scanning, CAD modeling adapted to the patient's anatomy, structural optimization with topological networks, and 3D printing with differentiated infill.
- Clinical practice: contact areas are optimized for comfort, while fastening systems (e.g., zip-tie) represent one of the major challenges.
- Vibrations are not yet part of clinical practice, but there is interest in integrating active modules for neuro-motor rehabilitation.
- Future perspectives aim at moving from plaster to plastic, followed by the integration of sensors and smart components.

Figure 19. Various orthosis models manufactured through 3D printing [99]

Following the numerical simulations, experimental testing, and practical documentation, the overall consistency of the results was confirmed, with the first natural modes being consistently identified in the low-frequency range, directly relevant from a clinical perspective. The differences between methods can be explained by the specific characteristics of 3D-printed materials and the simulation assumptions; however, the trends were mutually validated.

More importantly, the practical experience demonstrated that beyond numerical or experimental rigor, personalization, comfort, and clinical adaptability remain the decisive factors for the effectiveness of modern orthoses, representing the main applied contribution of this chapter.

CONCLUSIONS AND PERSPECTIVES

The purpose of this thesis was to explore the potential of mechanical vibrations as a therapeutic stimulation method in biomechanical rehabilitation, by integrating them into modern orthoses manufactured through 3D printing. The research combined theoretical analysis and literature studies with numerical simulations using FEM methods and physical tests on real prototypes, complemented by practical documentation in a professional setting. This integrated approach provided a balanced overview of the subject and confirmed the feasibility of applying vibrations in this field.

The results showed that, both through simulations and experimental tests, the natural vibration modes of AFO orthoses manufactured from PLA and PETG could be identified. The dominant frequencies were in the 20–40 Hz range, which confirms the initial hypothesis and supports the applicability of this frequency range in clinical settings. In addition, the analysis performed on test specimens demonstrated good agreement between simulations and measurements, validating the usefulness of FEM methods in the design and optimization of orthoses.

The original contributions of this work are reflected in several directions: the comparison between two different simulation platforms (FEBio and SolidWorks), the updated synthesis of the specialized literature, the performance of physical tests correlated with simulations, the proposal of an alternative analysis method through mechanical impulse, and the inclusion of practical documentation directly from the industrial environment. Together, these provide a comprehensive approach that combines theory, computer simulation, and applied experience in the design of modern orthoses.

However, the research also presents some limitations: FEM models assumed idealized materials, different from the layered structure of 3D-printed parts, and the physical tests did not perfectly reproduce the virtual conditions. Moreover, the study was limited to passive prototypes, without integrating active vibration modules. Future research directions aim at developing active orthoses with miniaturized vibration sources, creating a clinical testing protocol, and extending studies to other types of orthoses. Thus, this work may serve as a solid foundation for future intelligent orthotic devices, scientifically validated and adapted to the real needs of rehabilitation.

Selective Bibliography

[1]Organizația Mondială a Sănătății, Banca Mondială. *Raportul mondial privind dizabilitatea*. Geneva: OMS, 2011.

[2] Dollar, A. M., & Herr, H. (2008). *Lower extremity exoskeletons and active orthoses: Challenges and state-of-the-art*. IEEE Transactions on Robotics, 24(1), 144–158. https://doi.org/10.1109/TRO.2008.915453

[3] Rubin, C., Pope, M., Fritton, J. C., Magnusson, M., Hansson, T., & McLeod, K. (2003). *Transmissibility of 15-hertz to 35-hertz vibrations to the human hip and lumbar spine: Determining the physiological feasibility of delivering low-level anabolic mechanical stimuli to skeletal regions at risk of osteoporotic fracture*. Spine, 28(25), 2621–2627. https://doi.org/10.1097/01.brs.0000102682.61791.c9

[4] Rittweger, J. (2010). *Vibration as an exercise modality: how it may work, and what its potential might be*. European Journal of Applied Physiology, 108(5), 877–904. https://doi.org/10.1007/s00421-009-1303-3

[5] Cardinale, M., & Bosco, C. (2003). *The use of vibration as an exercise intervention*. European Journal of Applied Physiology, 91(1), 79–87. https://doi.org/10.1097/00003677-200301000-00002

[6] Kawanabe, K., Kawashima, A., Sashimoto, I., Takeda, T., Sato, Y., & Iwamoto, J. (2007). *Effect of whole-body vibration exercise and muscle strengthening, balance, and walking exercises on walking ability in the elderly*. Keio Journal of Medicine, 56(1), 28–33. https://doi.org/10.2302/kjm.56.28

[7] Bosco, C., Cardinale, M., & Tsarpela, O. (1999). *Influence of vibration on mechanical power and electromyogram activity in human arm flexor muscles*. European Journal of Applied Physiology and Occupational Physiology, 79(4), 306–311. https://doi.org/10.1007/s004210050512

[8] Bovenzi, M. (2006). *Health effects of mechanical vibration*. Giornale Italiano di Medicina del Lavoro ed Ergonomia, 28(1 Suppl), 58–64.

[9] Issurin, V. B. (2005). *Vibrations and their applications in sport: A review*. Journal of Sports Medicine and Physical Fitness, 45(3), 324–336

[10] Luo, J., McNamara, B., & Moran, K. (2005). *The use of vibration training to enhance muscle strength and power*. Sports Medicine, 35(1), 23–41. https://doi.org/10.2165/00007256-200535010-00003

[11] International Organization for Standardization. (2001). *ISO 2631-1: Mechanical vibration and shock — Evaluation of human exposure to whole-body vibration*. Geneva: ISO.

[13] Pasca, O. M. (2020). *Studii și cercetări privind ortezele de gleznă și picior în recuperarea neuromusculară* [Teză de doctorat]. Universitatea Tehnică din Cluj-Napoca.

[14] Megalingam, R. K., Manoharan, S. K., Mohandas, S. M., Reddy, C. P. K., Vijay, E., & Soman, K. P. (2023). *Wearable Hand Orthotic Device for Rehabilitation: Hand Therapy*

with Multi-Mode Control and Real-Time Feedback. Applied Sciences, 13(6), 3976. https://doi.org/10.3390/app13063976

[15] Nigg, B. M., & Herzog, W. (2007). *Biomechanics of the Musculo-skeletal System* (3rd ed.). Wiley.

[17] Hall, S. J. (2014). *Basic Biomechanics* (7th ed.). McGraw-Hill Education.

[18] Michael, J. W., & Bowker, J. H. (2002). *Atlas of Orthotics: Biomechanical Principles and Application* (3rd ed.). Mosby.

[20] Kluess, D., Wieding, J., Souffrant, R., Mittelmeier, W., & Bader, R. (2010). *Finite Element Analysis in Orthopaedic Biomechanics*. InTechOpen.

[21] Wittek, A., Miller, B., & Miller, K. (2010). Computational Biomechanics for Medicine. Springer.

[23] Webster, J. B., & Murphy, D. P. (eds.). (2019). *Atlas of Orthoses and Assistive Devices* (5th ed.). Elsevier.

[24] Perry, J., & Burnfield, J. M. (2010). *Gait Analysis: Normal and Pathological Function* (2nd ed.), ISBN: 978- 1556427664.

[25] Arus, E. (2012). *Biomechanics of Human Motion: Applications in the Martial Arts*. CRC Press.

[26] Rapoarte de cercetare proprii elaborate în cadrul studiilor doctorale.

[33] Orthotix. *Pavlik Harness – Paediatric Hip Dysplasia Orthosis*. Disponibil la: https://www.orthotix.co.uk/shop/paediatric/pavlik-harness/ [Accesat la: 6 aprilie 2025].

[38] Nascimento, L. M. S. do, Bonfati, L. V., Freitas, M. L. B., Mendes Junior, J. J. A., Siqueira, H. V., & Stevan Jr., S. L. (2020). *Sensors and Systems for Physical Rehabilitation and Health Monitoring—A Review.* Sensors, 20(15), 4063. https://doi.org/10.3390/s20154063

[39] Kim, Y., & Oh, K. (2023). *Cycling knee brace design analysis using 3D virtual clothing program to assess clothing pressure distribution and variance*. *Fashion and Textiles*, 10(1), 3. https://doi.org/10.1186/s40691-023-00354-8

[42] Klein-Nulend, J., Bacabac, R.G., & Mullender, M.G. (2005). *Mechanobiology of bone tissue.* Pathophysiology, 12(2), 85–95. https://doi.org/10.1016/j.patbio.2004.12.005

[44] Health and Safety Executive (HSE). *EU Good Practice Guide on Whole-Body Vibration. Version V6.7*, 2006.

[50] ISO 5349-1:2001. Mechanical vibration — Measurement and evaluation of human exposure to hand-transmitted vibration. Geneva: ISO.

[57] Raise3D. *A Guide to the Stages of 3D Printed Production of Orthopedic Insoles*. Available online: https://www.raise3d.com/case/a-guide-to-the-stages-of-3d-printed-production-of-orthopedic-insoles/ [accesat în februarie 2025].

[58] Ciobanu, O., Soydan, Y., Hızal, S. *Customized foot orthosis manufactured with 3D printers*. Proceedings of the International Conference on Manufacturing Systems, 2012.

[59] Xu, J., Bao, T., Lee, U. H., Kinnaird, C., Carender, W., Huang, Y., Sienko, K. H., & Shull, P. B. (2017). Configurable, wearable sensing and vibrotactile feedback system for real-time postural balance and gait training: proof-of-concept. *Journal of NeuroEngineering and Rehabilitation*, 14(1), 102. https://doi.org/10.1186/s12984-017-0313-3

[61] Bioness Inc. L300 Go Foot Drop System. Disponibil online la: https://bionessrehab.com/products/l300-go/ (accesat în aprilie 2025).

[62] Bethoux, F., Rogers, H. L., Nolan, K. J. și colaboratorii săi (2014). *The effects of peroneal nerve functional electrical stimulation versus ankle-foot orthosis in patients with chronic stroke: a randomized controlled trial.* Neurorehabilitation and Neural Repair, 28(7), 688–697. https://doi.org/10.1177/1545968314521007

[65] Farrell, T. R., & Weir, R. F. (2007). *The optimal controller delay for myoelectric prostheses.* IEEE Transactions on Neural Systems and Rehabilitation Engineering, 15(1), 111–118. https://doi.org/10.1109/TNSRE.2007.891391

[72] Lee, J., Kwon, K., Soltis, I., Matthews, J., Lee, Y. J., Kim, H., Romero, L., Zavanelli, N., Kwon, Y., Kwon, S., Lee, J., Na, Y., Lee, S. H., Yu, K. J., Shinohara, M., Hammond, F. L., & Yeo, W.-H. (2024). *Intelligent upper-limb exoskeleton integrated with soft bioelectronics.* Nature Electronics, 6(1), 105–114. https://doi.org/10.1038/s41528-024-00297-0

[75] Wong, Y. și colaboratorii săi (2023). *Upper limb practice with a dynamic hand orthosis to improve arm and hand function in people after stroke: a feasibility study*. Pilot and Feasibility Studies, 9, 132. https://doi.org/10.1186/s40814-023-01353-8

[79] Viceconti, M., Henney, A., & Morley-Fletcher, E. (2016). In silico Clinical Trials: how computer simulation will transform the biomedical industry. International Journal of Clinical Trials, 3(2), 37–46. https://doi.org/10.18203/2349-3259.ijct20161408

[80] Khan, F., Singh, K., & Carter, J. (2024). Vibration Behavior of 3D-Printed Graded Composites: Fabrication and Testing. *Polymers*, 16(23), 3428. https://doi.org/10.3390/polym16233428

[81] Umer, U., Mian, S.H., Moiduddin, K., & Alkhalefah, H. (2023). Exploring Orthosis Designs for 3D Printing Applying the Finite Element Approach: Study of Different Materials and Loading Conditions. *Journal of Disability Research*, 2(1), 85–97. https://doi.org/10.57197/JDR-2023-0011

[82] Brebbia, C. A., & Dominguez, J. (1994). *Boundary Elements: An Introductory Course*. Computational Mechanics Publications, ISBN 978-1-85312-349-8.

[83] Banerjee, P. K. (1994). *The Boundary Element Methods in Engineering*. McGraw-Hill. ISBN-13: 978-0077077693

[84] Yang, Y., & Kingan, M. (2024). A hybrid wave and finite element/boundary element method for predicting the vibroacoustic characteristics of finite-width complex structures. *Journal of Sound and Vibration, 582*, 118402. https://doi.org/10.1016/j.jsv.2024.118402

[85] Yao, L., Liu, G.R., & Nguyen-Thoi, T. (2017). A Coupled Smoothed Finite Element–Boundary Element Method for Structural–Acoustic Analysis of Shell. *Archives of Acoustics*, 42(1), 117–125. https://doi.org/10.1515/aoa-2017-0006

[86] Cundall, P. A., & Strack, O. D. L. (1979). A discrete numerical model for granular assemblies. *Geotechnique*, 29(1), 47–65. https://doi.org/10.1680/geot.1979.29.1.47

[89] Clough, R. W., & Penzien, J. (1993). *Dynamics of Structures* (2nd ed.). ISBN-10: 0070113947

[91] Steck, P., Scherb, D., Witzgall, C., Miehling, J., & Wartzack, S. (2023). *Design and Additive Manufacturing of a Passive Ankle–Foot Orthosis Incorporating Material Characterization for Fiber-Reinforced PETG-CF15*. Materials, 16(9), 3503. https://doi.org/10.3390/ma16093503

[92] Khantan, M., Avery, M., Aung, P.T. și colaboratorii săi (2023). The NuroSleeve, a user-centered 3D printed hybrid orthosis for individuals with upper extremity impairment. *Journal of NeuroEngineering and Rehabilitation*, 20:103. https://doi.org/10.1186/s12984-023-01228-2

[93] Chaneco. Basic Resting Hand Orthosis. Disponibil online la: https://www.chaneco.co.uk/product/basic-resting-hand-orthosis/

[99] H-Shift Engineering. *Design Guide – Hand Orthosis*. Document tehnic intern, transmis în cadrul colaborării pentru proiectul de cercetare doctorat