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Tissue Modeling and Simulation for Low-Intensity Vibration Devices

MASTER'S THESIS

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ABSTRACT

Low-intensity vibrations (LIV) are a promising technology in biomedical engineering, known for their potential to stimulate biological responses in soft tissues safely and non-invasively. This thesis focuses on developing computational and physical models to simulate the interaction of LIV with several tissues, aiming to understand their biomechanical effects. The study integrates finite element analysis (FEA) using ANSYS to evaluate tissue deformation, stress distribution, and potential resonance phenomena under different vibration parameters. Special attention is given to identifying safe frequency ranges that promote regenerative and therapeutic effects without risking mechanical damage. The outcomes of this work contribute to optimizing LIV-based therapies, providing a framework for future applications in physiology.

KEY WORDS

Lactation, Low-intensity vibrations, Biomechanical Effects

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1. Introduction

Many diseases and dysfunctions of the human body remain prevalent today, despite significant advancements in medical science. Conventional treatments often involve invasive procedures, high costs, and substantial risks, which can limit their accessibility and safety for broad patient populations. This way, the search for safer, non-invasive, and more affordable alternatives remains an important focus in modern healthcare research. In this context, biomedical devices are rapidly gaining importance as they offer innovative solutions for diagnosis, therapy, and rehabilitation.

Among these technologies, low-intensity vibration (LIV) devices have emerged as a promising therapeutic modality. However, the biomechanical response of tissues to LIV remains complex and highly dependent on tissue-specific mechanical characteristics, vibratory parameters (such as frequency), and individual patient factors. Given the ethical and practical challenges associated with extensive *in vivo* experimentation on human tissues, computational modeling has become a powerful tool for exploring and predicting tissue responses under LIV conditions. This thesis seeks to contribute to this field by developing computational models of biological tissues subjected to LIV.

1.1. Importance of Simulations and Tissue Modeling

Biomedical engineering is a recent area of research involved in biology and medicine, which is able to provide new designs and concepts for the diagnosis, treatment and prevention of several health complications (Zhang, 2015). One of the pillars of biomedical engineering is the creation of biomedical devices. These are defined by Aronson et al. (2020) as "a contrivance designed and manufactured for use in healthcare, and not solely medicinal or nutritional". Medical device-based therapies are playing an increasingly pivotal role in healthcare, expanding into global markets and bringing innovative technological solutions to the treatment and management of a wide range of diseases (Kramer et al., 2014).

In order to guarantee efficacy and security, various experiments must be carried out during the development phase of a biomedical device. Every medical device is made with a specific intended function, always keeping safety and performance in mind to ensure flawless operation (Altayyar, 2020). Biomedical researchers use different methods to conduct these experiments, including experimental animals, tissue, and cell cultures as well as computer-aided simulations and clinical studies (Saeidnia et al., 2015).

Animal models have been the cornerstone of biomedical research for decades. They provide invaluable insights into human disease mechanisms, preventative and therapeutic targets, and treatment strategies, making them indispensable tools for translational research (Chang and Grieder, 2024), including the development of a biomedical device.

Despite animal models having played a vital role in studying significant and growing diseases that impact human health worldwide (Domínguez-Oliva et al. 2023), lots of ethical questions still emerge when experimental setups that implicate animals are considered (Guimarães, 2025). Nonetheless, animal models have significant limitations in uncovering

many of the essential features of pathology and disease processes that are specific to humans (Zhao and Bhattacharyya, 2018). Due to this fact, humans are an irreplaceable resource in biological research, as experiments carried out upon them provide insights that no other model can totally mimic.

However, there are several obstacles to overcome when performing *in vivo* research on human beings, as clinical research is based on ethical standards that respect the study's participants, and protect both their health and rights (Lee, 2022). These obstacles include moral dilemmas, exorbitant expenses, logistical difficulties, and physiological response variability. The scope and frequency of experimental interventions are frequently limited by these constraints, which make such research challenging to plan and carry out.

Different from animal and human testing, tissue modeling provides a safe and morally harmless alternative that can provide powerful results and drive innovation, particularly in the first steps of biomedical device creation. The arrival of computers in the late 1990s triggered a chain reaction across various branches of engineering, with biomedical engineering being one of them. It transformed how both researchers and medical professionals tackled issues within healthcare systems (Shukla et al., 2024).

Models are simplified representations of complex processes or structures to aid in representing, analyzing, and understanding those same processes (Post et al., 2022). Tissue modeling refers then to the creation of mathematical and computational representations that depict biological tissues, allowing for the simulation of their structure, function, and mechanical properties.

Tissue modeling often involves the usage of techniques such as finite element analysis (FEA) to capture the unique properties of tissues. In the most recent years, a significant progress in understanding the biomechanical properties and structures of soft biological tissues has emerged, and is accompanied by developments in the constitutive modeling of many soft tissues (Holzapfel and Ogden, 2025). Precisely modeling the mechanics of biological tissues offers numerous benefits and assists researchers in various biomedical applications (Khaniki et al. 2023), including biomedical devices.

Models can represent individual cells (Goldberg et al., 2018), layers of tissue (Wenk et al., 2012), or even entire organs (Gerasimov et al., 2021), allowing researchers to explore how tissues respond to various internal and external stimuli, such as mechanical forces, injury, or disease. These models are built using experimental data obtained from *in vitro*, *in vivo*, or *ex vivo* studies, and can be tailored to specific tissue types, such as bone (Engelke et al., 2016), muscle (Stavness et al., 2012), cartilage (Anderson et al., 2008), or soft tissues like the breast, brain, or skin.

With accurate models, researchers can replicate disease progression (Xu et al., 2024), predict the results of surgical procedures (Delingette and Ayache, 2004), and design and refine biomedical devices (Mollica and Ambrosio, 2009). In addition, investigators are able to use models to systematically analyze systems perturbations and develop hypotheses to guide the design of new experimental tests (Fischer, 2008).

In recent years, advances in mathematical and computational sciences have led to new methodologies that significantly enhance our ability to predict *in vivo* tissue behavior (Post et al., 2022). These models are created using engineering materials. Unlike traditional engineering materials, biological tissues exhibit complex, nonlinear (Panda and Buist, 2018) and often viscoelastic behavior (Wu et al., 2022), which makes accurate modeling both challenging and essential.

In soft tissue research, models are used to understand pathologies such as pressure ulcers (Keenan et al., 2022), joint degeneration (Hellwig et al., 2016), and ligament injuries (Ren et al., 2022). Moreover, in the emerging field of mechanobiology, tissue simulations help explore how mechanical forces influence cellular behavior, contributing to insights into processes like wound healing (Ziraldó et al., 2013) and tumor progression (Yin et al., 2019).

Tissue modeling also holds a significant position in practical areas such as sports science and rehabilitation, where biomechanical models are used to evaluate injury risk, enhance athletic performance, and create efficient recovery protocols (Penichet-Tomas, 2024). In research and development, simulations allow for rapid iteration and feedback during device prototyping (Stahl, 2002) and therapy design, speeding up innovation cycles while lowering costs.

Importantly, incorporating tissue modeling into biomedical research (regardless of its inherent limitations) enhances both ethical and scientific standards. It encourages a more thoughtful and effective utilization of human tissue while minimizing reliance on animal models (Baker et al., 2016). Considering the ethical, financial, and logistical challenges associated with conducting *in vivo* studies on humans, simulation-based research presents a responsible alternative that can still provide high-quality, clinically relevant insights. As biomedical technologies persist in advancing, the necessity for accurate, validated tissue models will only increase, aiding in safer, more effective treatments and broadening our comprehension of intricate biological systems (Shukla et al., 2024).

1.2 LIV Devices

1.2.1 Basic Principles

LIV devices are non-invasive tools engineered to deliver gentle, high-frequency mechanical stimulation to biological tissues without causing substantial physical stress. These devices emit mechanical signals in the form of small, rapid oscillations, typically centered around 30 Hz, as supported by Khan et al. (2024). They operate at low amplitudes and maintain accelerations below 1g, making them suitable for repeated use, even in vulnerable populations. While 30 Hz is a commonly referenced standard, some research has reported effective applications at higher frequencies, ranging from 45 up to 90 Hz, depending on the therapeutic target and context (Roberts et al., 2021). Higher frequencies might still be useful in other clinical contexts, although common studies are lacking presenting this setup.

The fundamental concept behind LIV is that even slight mechanical inputs can trigger mechanosensitive structures in cells and tissues, initiating physiological responses via mechanotransduction pathways (Baskan et al., 2020). These devices are designed to mimic

the subtle vibratory signals that naturally occur within the body, fostering advantageous results such as enhanced circulation, as proved by Mahbub et al. (2019), where exposure to whole body vibration using frequencies ≤ 30 Hz provoked an improvement in peripheral blood flow, or amplified cellular communication, as demonstrated by Uzer et al. (2014).

In contrast to high-impact mechanical loading, which can be inappropriate for elderly or clinical populations, LIV provides a safer, milder alternative that has demonstrated potential in addressing osteoporosis, muscle atrophy, and mobility limitations. Numerous studies have shown that LIV can promote osteoblast and myocyte activity while inhibiting osteoclast-induced bone resorption, emphasizing its potential in preserving or even enhancing musculoskeletal integrity in at-risk groups.

1.2.2 Historical Context

In a review conducted by Kaeding (2016), the historical context of vibration therapy is examined. It is well-established that the use of vibrations for healing purposes is indeed a primordial practice. Details regarding the therapeutic use of vibrations trace back to ancient Greece and Rome. For instance, riding a horse or donkey and riding a two-wheeled carriage over uneven, rough stone roads were both associated with vibrations. The Greeks used saws wrapped in cotton fabric to direct vibrations to particular parts of the body. Lucius Annaeus Seneca added that carriage travel was good for maintaining good health.

During the 16th century, a Japanese document outlined the possible health benefits of vibrations. In this regard, the use of vibrations for alleviating rheumatic conditions and promoting the healing of broken bones was described. In 1808, John Barclay published the book "The Muscular Motion of the Human Body," in which he detailed a case of muscular spasm treated through vibrations. John Harvey Kellogg (1852-1943) is arguably the most renowned pioneer of modern vibration therapy. He invented vibrating chairs, bars, and platforms that are currently available on the market and employed them in his "Battle Creek Sanatorium" for therapeutic applications.

The initial models were driven by steam, and these machines facilitated the use of whole body vibrations. Currently, there is a diverse selection of devices that use different methods and mechanisms available on the worldwide market, with the majority employing sinusoidal vibrations. As a result, the spectrum of conditions for utilizing vibrations has broadened. Typically, devices that generate relatively high-frequency sinusoidal vibrations are employed in a large area.

1.2.3 Clinical Context

Several important examples are introduced here to give a glimpse of the relevance and impact of LIV across different fields, even though a thorough discussion of LIV applications will be presented in the state-of-the-art section of this work.

LIV's use has expanded dramatically in recent years, reaching fields like space medicine, sports science, and rehabilitation. LIV has been investigated as a way to improve

neuromuscular activation and facilitate post-exercise recovery, with LIV hastening the elimination of metabolic waste products, lessening soreness and speeding up recovery (Cheng et al., 2017). Similar to this, NASA has looked into LIV as a non-invasive way to prevent astronauts from losing bone and muscle during extended microgravity exposure (Thompson et al., 2020). LIV is becoming more and more relevant for sedentary populations outside of high-performance contexts. It provides a preventive approach to counteract the negative consequences of inactivity, like decreased bone density and poor circulation (Uzan et al., 2016).

Within healthcare, studies show that LIV can inhibit bone resorption (Xie et al., 2006), increase bone formation (Castillo et al., 2006; Xie et al., 2006), and thereby increase bone density and lower the risk of fractures (Thompson et al., 2014). Since LIV promotes muscle activation, reduces atrophy, and enhances local blood flow, its use in muscle rehabilitation has also shown promise. However, the circulatory benefits seem to be transient once the vibration stops (Tzen et al., 2018). Additionally, by increasing tendon elasticity and promoting collagen synthesis, LIV has shown promise in the treatment of soft tissue injuries (Thompson et al., 2015).

Depending on the intended use, LIV device design and implementation can vary greatly. The main goals of full-body platforms, which are frequently employed in clinical and rehabilitation settings, are to improve muscle coordination, proprioception, and bone strength by sending mechanical signals throughout the body. Conversely, localized wearable stimulators provide customized therapies without requiring full-body engagement by focusing on particular anatomical regions. The use of LIV in routine health management is being made possible by the growing trend of its integration into wearable and portable technologies.

Computational modeling plays an equally significant role in improving our capacity to model tissue responses and customize treatment regimens. LIV has the potential to be a key component of non-invasive therapy in a wide range of applications, from improving performance and recuperation in healthy populations to treating chronic conditions, as these technologies develop (Albasini et al., 2010).

1.2.4 Need For Simulation To Understand LIV Effects

Despite the promising benefits, research is still ongoing to determine the exact biomechanical and cellular processes by which LIV works. Determining the optimal frequency, amplitude, and duration of exposure for various tissue types is one of the challenges in optimizing LIV therapy because responses can differ depending on age, underlying medical conditions, and individual physiology.

Tissue modeling is thus indispensable for understanding the effects of LIV because it provides a detailed computational framework to analyze how biological tissues respond to mechanical stimuli at different scales, without the need of utilizing animal or human participants. LIV influences various tissue types, including bone, muscle, and connective tissues, but its effects depend on factors such as frequency, amplitude, exposure duration, and individual physiological characteristics. Since direct in vivo measurements of mechanical stress and strain in human tissues are difficult, computational models allow researchers to

simulate these interactions and predict how different tissue structures respond to LIV. By using finite element modeling (FEM) and other advanced simulation techniques, scientists can investigate the mechanical deformation of tissues, assess stress distribution, and identify optimal vibration parameters for therapeutic applications.

Furthermore, tissue modeling allows for the exploration of LIV effects on different biological structures, highlighting variations in mechanical properties across tissues and identifying which areas are most responsive to vibration stimuli. By integrating experimental data with computational simulations, researchers can develop tissue-specific models that account for these complexities and improve the precision of LIV applications. In rehabilitation and biomechanics, these models help determine the safest and most effective vibration protocols for individuals with varying health conditions, such as aging populations or patients recovering from musculoskeletal injuries. As LIV technology advances, computational simulations will remain essential in guiding its clinical applications, refining device designs, and ensuring that vibration-based therapies are tailored to meet specific patient needs.

1.3 The Breast And Lactation

An organ associated with mammals, the breast is essential for raising the young. Any member of a group of vertebrate animals whose offspring are fed milk from the mother's breast (mammary glands) is referred to as a mammal. This way, one of the main characteristics of the mammalian class is lactation, which is facilitated by the breast. (Semakane et al. 2024). Lactation is thus a complex physiological function in mammals that serves an essential purpose in delivering early nutrition and immune defense to offspring (Power and Schulkin, 2013).

Human lactation is controlled by a complex interplay between mechanical stimulation, tissue remodeling, and hormonal signals (Macias and Hinck, 2012). Due to the obvious importance of the breast in the lactation process, an anatomical explanation is prompted. A radial arrangement of 15 to 20 lobes makes up each breast, and each lobe is further subdivided into lobules, which are composed of groups of alveoli (Khan and Sajjad, 2019). Encircled by epithelial cells that produce and secrete milk, these alveoli serve as the milk-producing units (Oakes et al., 2006). Encircling the alveoli are myoepithelial cells that contract, pushing milk through a system of ducts toward the nipple (Biswas et al., 2022). Underneath the areola, the ducts unite to form lactiferous sinuses, which temporarily store milk before it is expelled. Suspensory ligaments, also known as Cooper's ligaments, are fibrous bands that securely attach the breast to the skin and underlying structures. These bands serve to support, hold, and shape the breasts (Ramião et al. 2016).

Regarding the development of the breast, it's around the seventh or eighth week of pregnancy when tissue invades the stroma to form the mammary disc, marking the beginning of mammary gland development (Geddes, 2007). The interactions between the mesenchyme and epithelium, which start during embryonic development, also direct the mammary gland's growth and maturation (Hynes and Watson, 2010). Smooth muscle in the nipple and areola forms between weeks 12 and 16, while epithelial outgrowths and parenchymal branching take place between weeks 10 and 20. Solid epithelial cords begin to form at 20 weeks, and by 32 weeks, they have canalized into primary milk ducts that open onto the growing nipple. Connective tissue gives rise to adipose tissue, which is necessary for additional glandular

development. During lactation, the breast reaches its full functional maturity (Geddes, 2007). In Figure 1, the main components of the human breast may be observed.

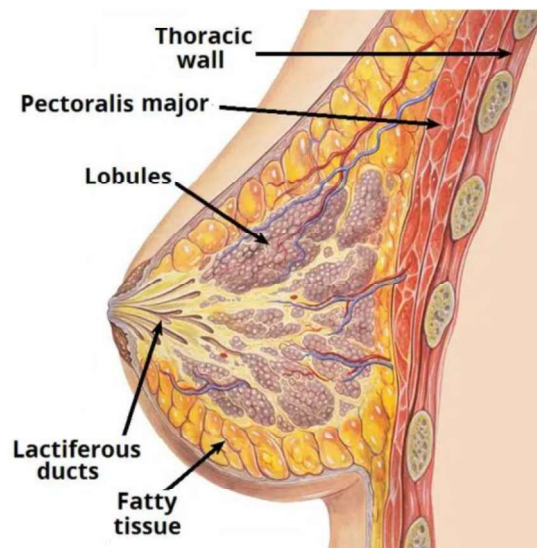


Figure 1. Human breast anatomy. Image retrieved from TeachMeAnatomy (n.d.).

Endocrine and neuroendocrine pathways tightly regulate the physiology of lactation. While oxytocin, which is released from the posterior pituitary in response to nipple stimulation, promotes milk letdown, prolactin, which is released from the anterior pituitary, stimulates milk synthesis (Crowley, 2015). Progesterone and estrogen levels must fall after delivery in order to facilitate efficient lactogenesis, even though they increase during pregnancy and encourage ductal and lobular growth (Slepicka et al., 2021). The suckling reflex mechanically causes milk ejection by inducing rhythmic contractions of the myoepithelial cells and stimulating the neurohormonal release of oxytocin (Crowley, 2015).

Lactation cannot continue unless milk is regularly removed after delivery because this causes the neurohormonal release of prolactin and oxytocin. Mechanical stimulation, which is usually accomplished through infant suckling but can also be achieved with breast pumps, is necessary to maintain milk flow and avoid milk stasis (Zinaman et al., 1992). Milk ejection is thus critical for successful lactation. Failure to remove sufficient quantities of milk results in a decrease in milk production because of the local control mechanisms (Geddes 2007).

Because lactation is an intrinsically complex physiological phenomena, it does not come without complications. Breastfeeding-related issues and diseases affect up to 50% of nursing mothers and are the most common cause of early breastfeeding termination (Schwartz et al., 2002). As a result, many women encounter issues that hinder breastfeeding, including clogged ducts, nipple problems (such as abrasions, cracks, and ulcerations), milk stasis, and, most importantly, mastitis, an inflammatory disease of the breast that can be excruciating and incapacitating (Abou-Dakn et al., 2010). Milk stasis and bacterial infections frequently cause mastitis, which manifests as swelling, redness, and systemic symptoms like fever and exhaustion. For instance, before realizing that they have breast tenderness or a blocked duct, women with mastitis frequently have influenza-like symptoms (Geddes, 2007). It may develop into an abscess if treatment is not received, necessitating more extensive medical care (Blackmon and Mukherji, 2020). Because lactation plays a vital role in both the short-term and

long-term health of infants and also shields mothers from breast cancer, ovarian cancer, and type 2 diabetes mellitus (Hannan et al., 2023), new therapeutic strategies that target these complications would be of great importance.

In this regard, it is hypothesized that LIV may provide therapeutic advantages in breast tissue to treat lactation-related complications like mastitis and milk stasis. It is thought that by gently stimulating breast tissue, LIV may enhance lymphatic drainage, promote local circulation, and reduce inflammation, elements that are commonly linked to the onset of mastitis. Improved lymphatic flow may aid in infection prevention and obstruction relief by removing cellular debris and trapped milk from the ducts. Additionally, a study by Corbiere and Koh (2020) showed that local LIV at a frequency of 90 Hz enhances the healing of muscle injury in mice, suggesting that LIV may support tissue repair. By enhancing microcirculation, LIV may also change the local immune response, which can aid in the prevention and resolution of inflammation. This is supported by a study from Blanks et al. (2020), where it was found that whole body vibration elicits differential immune and metabolic responses in obese and normal weight individuals. Despite the lack of direct research on LIV and lactation-related disorders in humans, the benefits it has been shown to have for soft tissue healing provide a promising basis for further study. This raises the possibility of developing non-invasive, supplemental treatments for breastfeeding-related problems that are currently managed with massage, manual drainage, or antibiotics.

Because simulation offers a safe environment for analyzing biomechanical interactions inside breast tissue, it is crucial for investigating the potential effects of LIV on lactation-related problems. Computational modeling methods such as finite element analysis (FEA) can be used to generate anatomically informed models of the breast in order to predict the effects of mechanical vibrations on tissue-level responses. These models can identify regions where mechanical stimulation may enhance blood flow and lymphatic drainage by predicting patterns of stress within the tissue. Furthermore, simulation can aid in approximating the impact of vibration on fluid flow through the milk ducts, which is especially important when discussing milk stasis and blockage. By altering parameters like the frequency, assessing how LIV parameters affect physiological processes like drainage, flow dynamics, and inflammatory responses is possible without the risks associated with in vivo testing.

1.4 Bone And Joints

The benefits of LIV expands far beyond soft tissues like the breast. For instance, its role in altering the physiology of bones and joints is non-negligible, and will be investigated in this section.

Bone is a highly specialized form of connective tissue (Mohamed, 2008) that serves multiple critical functions in the human body, including providing structural support (Zimmermann and Ritchie, 2015), protecting vital organs (Cowan et al., 2024), enabling movement through articulation with muscles (Dermience et al., 2015), and serving as a reservoir for minerals such as calcium and phosphate (Copp and Shim, 1963). Bone tissue is composed of two primary types: cortical bone and trabecular bone. Cortical bone, also known as compact bone, forms the dense outer layer of bones and accounts for approximately 80% of skeletal mass. It provides strength and rigidity to the structure. Trabecular bone, also called cancellous or spongy bone, is found inside bones at the ends of long bones and within

vertebrae, characterized by a porous, lattice-like structure that contributes to energy absorption and metabolic activity (Clarke, 2008).

In terms of physiology, bone is a highly vascularized, dynamic tissue that can undergo continuous remodeling over the course of a lifetime. By coordinating the activities of osteoclasts, which resorb bone, and osteoblasts, which create new bone matrix, this remodeling process enables bones to adjust to mechanical demands, fix microdamage, and maintain mineral homeostasis (Clarke, 2008). Through mechanotransduction pathways, which convert external forces into cellular reactions that either promote bone formation or inhibit resorption, mechanical loading is essential for preserving bone health. Bone density, strength, and structural integrity are ultimately determined throughout the lifespan by the complex interplay among mechanical stimuli, hormonal regulation, and nutritional status (Robling and Turner, 2009).

Age-related chronic inflammation is a common denominator of osteoporosis, a condition that significantly contributes to disability and frailty (Greco et al., 2019). Due to improvements in our knowledge of the signals that control bone formation and resorption, osteoporosis, a skeletal disorder marked by low bone mineral density and an increased risk of fracture, has been thoroughly studied (Bilezikian et al., 2009). In the majority of postmenopausal women with osteoporosis, bone loss is associated with estrogen deficiency. Rapid bone loss results from an imbalance between bone resorption and bone formation as well as an increase in bone turnover (Cheng et al., 2022).

The risk of developing osteoporosis is higher under conditions including aging (Pietschmann et al., 2009), hormonal deficiencies (Cheng et al., 2022), chronic inflammation (Greco et al., 2019), and the absence of mechanical loading, such as in cases of immobility, chronic bed rest, or prolonged spaceflight (Carina et al., 2020).

The skeletal system can move, be stable, and be flexible thanks to joints, which are specialized anatomical structures where two or more bones meet. From highly mobile synovial joints, like the knee and shoulder, to immobile fibrous joints, like those in the skull, they are categorized according to their structure and function. The most prevalent kind of joints are synovial joints, which are distinguished by the presence of articular cartilage covering the ends of the bones, a joint cavity filled with lubricating synovial fluid, and a supporting joint capsule (Juneja et al., 2024).

Joints are extremely vulnerable to a variety of diseases and degenerative conditions, despite their vital role in mobility. For example, degenerative joint disease, commonly referred to as knee osteoarthritis, is usually brought on by articular cartilage deterioration and progressive loss (Hsu and Siwiec, 2023). Pain, inflammation, decreased range of motion, and a marked decline in quality of life are all possible outcomes of this condition. The importance of subchondral bone remodeling in the onset of osteoarthritis is becoming more widely recognized, and it's becoming a therapeutic target (Zhu et al., 2021).

LIV has shown growing promise as a non-invasive intervention to mitigate the progression of osteoporosis (Wysocki et al., 2011). In bone, LIV has been associated with increased bone formation, improved trabecular architecture, and suppression of bone resorption markers. By controlling the activity of cells involved in bone remodeling and influencing the differentiation fate of their mesenchymal and hematopoietic stem cell

progenitors, LIV enhances bone quality, according to mounting preclinical and clinical data (Chan et al., 2013).

Theoretically, LIV could be used to control subchondral bone remodeling and thereby slow the advancement of osteoarthritis in the knee. Literature of LIV usage in joint complications is still scarce. However, following the suggestion of whole-body vibration (WBV) as a therapeutic instrument for the prevention of subchondral bone loss in knee osteoarthritis, a study by Junbo et al. (2017) was conducted. There, the effects of different frequencies of WBV on subchondral trabecular bone microarchitecture, cartilage degradation and metabolism of the tibia and femoral condyle bone were assessed. It was found that at lower frequencies (10 Hz and 20 Hz) of low-magnitude WBV, bone turnover increased, cartilage degeneration was delayed, and a significant functional change of the rabbit model was caused, but osteoarthritis progression was not reversed after 8 weeks of treatment.

When assessing how LIV affects bones and joints, simulation plays a critical role. The observation of resonant effects as well as the tissue stress and deformation brought on by various frequencies is made possible by computational modeling. Finite element simulations are used to forecast stress distributions in bone that affect osteogenic activity.

1.5 Ovary

The term ovary references the female gonad. Both ovaries are a pair of intraperitoneal endocrine organs that are normally located in the lower left and right abdominal quadrants, respectively. The ovaries are two almond-shaped organs that are structurally part of the pelvic cavity and are connected to the uterus by the ovarian ligaments. The first function of the ovary is hormone production, which changes at puberty. The ovary's granulosa and theca cells secrete a variety of hormones. As gonadotropin-releasing hormone (GnRH) levels rise, the ovaries will start secreting more estrogen, testosterone, inhibin, and progesterone (Gibson and Mahdy, 2023).

Numerous health issues can affect the ovary, which can have a major effect on both reproductive function and general health. Common ovarian disorders include polycystic ovary syndrome (PCOS), which affects roughly 5–18% of individuals with ovaries, and is a metabolic, reproductive, and psychological condition which makes an impact throughout life. The cause is multifaceted and includes mechanisms linked to obesity, excess androgen exposure, insulin resistance, hypothalamic and ovarian dysfunction, and genetic and epigenetic susceptibility (Joham et al., 2022). Other illnesses impacting the ovary include ovarian cysts, which are typically benign fluid-filled sacs that can cause pelvic pain, bloating, and complications such as rupture or torsion, potentially cutting off blood supply (Mobeen and Apostol, 2023).

LIV applications in relation to ovarian health have not yet been explored. Future research on the ovary's mechanical reaction to vibratory stimulation is necessary, given the variety of conditions for which LIV has shown therapeutic promise. By improving microcirculation and lymphatic drainage, LIV may have potential for reducing ovarian conditions like ovarian cysts or PCOS. This thesis will model how a healthy ovary would react to sinusoidal mechanical loading in order to investigate this possibility. Understanding how

ovarian tissue behaves mechanically under LIV may help create new, non-invasive treatment approaches that support ovarian function and treat particular gynecological disorders.

1.6 Finite Element Analysis

Because biological tissues have complex geometries and material properties, it is frequently infeasible to solve the governing differential equations analytically. To bypass this complication, solutions are often approximated using numerical techniques such as the finite element method (FEM) (Freutel et al., 2014). In FEM, a continuous system described by partial differential equations is discretized into a finite number of algebraic equations. This process, also known as reducing a continuum model to a lumped-parameter model, simplifies the intricate structure of soft tissues into a mesh of finite elements. Each component simulates local behavior, and together they replicate the overall tissue response under physiological conditions. The efficiency of FEM in simulating tissue deformation, stress distribution, and vibrational behavior in biomedical applications is based on this fundamental idea (Bathe, 2006).

Stress and deflection inside a structure can be predicted using FEA. The model in FEA is separated into a grid of elements, which are basic shapes for which the program can generate governing equations in the form of a stiffness matrix. The displacements at the node points, where elements are connected, are the unknowns in each element. The stiffness matrix for the entire structure is put together and solved under the specified boundary conditions, producing node displacements from which element stresses can be computed (Raju et al., 2015). FEM is particularly prevalent in biomechanics for modeling soft tissue deformation, and the computational load depends on mesh complexity and problem scale (Martínez-Martínez et al., 2017).

Constitutive equations, which explain the stress-strain relationships of the modeled tissue, define the material behavior in FEM. Although many common materials can be accurately described by simple idealized models like the Newtonian viscous fluid or Hookean elastic solid, biological tissues usually display complex, nonlinear, and viscoelastic behaviors that resist such simplification (Fung, 2013). Accurate modeling of biological materials thus requires advanced constitutive frameworks.

The ability of FE models to predict in vivo behaviour strongly depends on the accuracy of the mechanical properties of tissue components. Setting the breast as an example, an accurate breast model has proved to be very difficult to implement, due to several difficulties associated with breast tissues. These include the breast's complex morphology, the patient-specific variability, the highly nonlinear mechanical behaviour, and the difficulty of measuring elastic properties of different types of tissues in the breast (Ramião et al., 2016).

Understanding how any tissue reacts dynamically to oscillatory stimuli is essential for researching the mechanical effects of LIV on that same tissue. To determine the tissue's inherent frequencies and associated mode shapes, modal analysis is utilized. These characteristics show how the structure naturally vibrates and are essential for avoiding or taking advantage of resonance phenomena, which occur when an external vibration close to a natural frequency significantly increases internal stress and deformation (Fu and He, 2001; Ramesha et al., 2015). Building on this, harmonic analysis replicates the cyclic nature of LIV

by modeling the tissue's steady-state reaction to continuous sinusoidal excitation (Kohnke, 1982). By observing tissue deformation, stress accumulation, and vibrational energy distribution across frequencies, evaluating mechanical safety, predicting therapeutic efficacy, and optimizing device parameters is possible.

While FEM is powerful for analyzing tissue deformation, stress, and vibration, computational fluid dynamics (CFD) offers complementary insights, especially in lactation research. CFD provides a numerical approximation to the equations that govern fluid motion (Zawawi et al., 2018). CFD enables the simulation of milk flow through the ductal system, predicting velocity fields, as milk ejection is a fluid transport process.

2. Objectives

This thesis's main objective is to generate computational models that simulate how different biological tissues react mechanically to LIV in order to support non-invasive treatment approaches in the biomedical setting.

The following particular goals have been set in order to fulfill the overall purpose of this study:

1. Model Development

- Develop tissue-specific finite element models that capture the mechanical behavior of soft and hard tissues under LIV loading.
- Incorporate appropriate material parameters and shapes to accurately reflect the nature of biological tissues according to existing literature.

2. Simulation of LIV Conditions

- Simulate the tissue response to different LIV frequencies.
- Apply modal and harmonic analyses to identify resonant frequencies and vibration modes relevant to each tissue type.

3. Tissue-Specific Applications

- Breast tissue: Model the impact of LIV on breast tissue, as well as milk's flow through the mammary duct.
- Ovarian tissue: Simulate the mechanical behavior of ovarian structures to explore potential therapeutic implications.
- Bones: Simulate the application of LIV into a human femur.
- Joints: Investigate the biomechanical response of the bony parts of the knee.

3. State-of-the-art

An overview of the most significant advancements in tissue simulations and LIV applications is given in this chapter. The biomechanical characterization of biological tissues, the development of simulation techniques like FEA and CFD, and the expanding use of mechanical stimuli in clinical and rehabilitation settings will be the main topics of discussion. This section attempts to highlight current knowledge, identify current limitations, and define the research landscape in which this thesis is situated by looking at both foundational and cutting-edge studies.

3.1. LIV Applications

To investigate the current uses of LIV in a variety of domains, such as clinical practice, space travel and even animal care, a comprehensive literature review was conducted.

LIV has been investigated as a possible way to affect the physiological functions of nursing animals, especially dairy cows. Sederevičius et al. (2023) looked into how LIV affected the udder and general milk production metrics. The results showed that LIV increased blood flow in the udder, which may have advantages for better tissue perfusion and circulation. The study did not find any significant changes in the main parameters of milk production, such as milk yield, composition, and quality indicators like somatic cell count and fat content, in spite of this physiological reaction. Furthermore, there were no discernible impacts on animal welfare, indicating that the cows were not subjected to any stress or discomfort by LIV. These results imply that although LIV might improve some physiological reactions, its direct effect on the efficiency of milk production is still unclear and needs more research to identify its possible uses in dairy farming.

LIV has shown promising potential for improving wound healing, particularly in diabetic conditions where healing is often impaired. Weinheimer-Haus et al. (2014) found that LIV boosted angiogenesis, granulation tissue formation, and wound closure in diabetic mice, while reducing harmful inflammation and increasing pro-healing factors. Building on this, Roberts et al. (2021) showed that low-frequency, low-acceleration LIV (45 Hz, 0.3 g) was the most effective protocol, consistently enhancing healing outcomes. These findings suggest that, with the right parameters, LIV could be a useful non-invasive therapy for chronic wounds, especially in diabetic patients.

Fu et al. (2021) explored the potential of low-magnitude high-frequency vibration (LMHFV) as a treatment for diabetes-induced bone loss, looking at both in vivo and cellular effects. They found that LMHFV (35 Hz, 0.25 g, 15 min/day) countered the negative impact of high glucose on osteoblast migration, proliferation, and differentiation by activating the GSK3 β / β -catenin pathway and boosting osteogenic markers like Runx2 and Cyclin D1. In diabetic rats, LMHFV improved bone strength, microarchitecture, and mineral composition, restoring key measures like femoral stiffness, trabecular density, and bone crystallinity. These

results highlight LMHFV's promise as a non-invasive strategy to improve bone health and reduce the damaging effects of diabetes on bone tissue.

Newberg et al. (2020) studied how isolated cell nuclei respond to LIV, revealing new insights into nuclear mechanosensitivity. Using atomic force microscopy, they found that mesenchymal stem cell nuclei stiffen by 75% after LIV (0.7 g, 90 Hz, four 20-minute bouts), even without changes in nuclear envelope proteins like Sun-2 or LaminA/C. This stiffening depended on the LINC complex and was linked to a 25% reduction in the chromatin-to-nuclear area ratio, suggesting chromatin compaction as the key driver. Importantly, the study showed for the first time that isolated nuclei can functionally respond to mechanical stimuli, deepening our understanding of how physical forces influence gene regulation and cell behavior.

Thompson, Yen, and Rubin (2014) reviewed the growing use of LIV therapy as a clinical tool for improving bone and muscle health, especially in people who can't engage in regular physical activity. They highlight that mechanical loading is key to maintaining skeletal strength and that LIV can mimic the bone-building effects of exercise without drugs. Studies show that LIV (<1g) promotes bone formation, reduces bone loss, and encourages stem cells to develop into bone-forming cells. It may also help combat sarcopenia, reducing frailty and fall risk in older adults. The authors stress the importance of carefully tailoring vibration protocols to patient needs and support using LIV alongside other treatments for bone-related conditions.

The potential of whole-body vibration (WBV), especially low-magnitude high-frequency vibration (LMHFV), to promote fracture healing has been thoroughly investigated in pre-clinical models. Through a systematic review of 19 pre-clinical studies involving different animal models and fracture types, Wang et al. (2017) determined that frequencies between 35 and 50 Hz and magnitudes under 1g were linked to generally favorable outcomes, including increased bone mineral density, vascularization at the fracture site, and callus formation. Interestingly, LMHFV seemed to work especially well in ovariectomized (OVX) animals, mimicking the osteoporotic conditions that are typical in postmenopausal women. The review also points out discrepancies, suggesting that the therapeutic window for frequency and intensity is still limited and not always effective. While the majority of studies reported positive effects, others found no discernible changes, and one study even noted impaired healing at 45 Hz (Wehrle et al., 2014).

The application of WBV to clinical practice is still in its infancy despite its potential. The review's studies were all pre-clinical; no human trials were discovered, and the ability to perform a meta-analysis was restricted by a high level of methodological heterogeneity, which included variations in animal species, fracture models, and vibration regimes. Furthermore, the type of fracture and fixation technique appear to affect the efficacy of WBV: rigid fixation appeared to lessen WBV sensitivity, and diaphyseal fractures responded better than metaphyseal ones. The review shows that additional systematic, large-scale clinical trials are necessary to confirm efficacy and optimize treatment protocols for human fracture healing, even though LMHFV is generally regarded as safe with no unintended side effects reported (Wang et al., 2017).

LIV has shown promise as a non-drug approach for preventing and treating osteoporosis. Studies show that low-magnitude, high-frequency mechanical signals (<1g; 20–

90 Hz) promote bone formation, reduce bone breakdown, and improve bone density and strength. Unlike traditional loading that relies on high tissue strain, LIV works through acceleration-driven effects that influence cellular activity, including stem cell differentiation toward bone-building pathways. Importantly, LIV meets ISO safety standards, making it suitable for vulnerable groups like the elderly or those with limited mobility. Overall, LIV offers a safe and effective mechanical alternative or complement to drug-based osteoporosis treatments (Chan et al., 2013).

Corbiere and Koh (2020) studied the effects of locally applied LIV on muscle regeneration after injury in mice. They found that applying LIV (90 Hz, 0.2 g, 30 min/day) significantly boosted muscle fiber growth and reduced tissue damage compared to controls, without increasing IGF-1 levels. Instead, LIV's benefits were linked to reduced expression of atrophy-related genes (FOXO1, Atrogin-1) and increased Akt signaling. In cultured muscle cells, LIV also improved myotube fusion and size, again tied to Akt activation. These findings suggest that LIV supports muscle repair through local cellular pathways, offering mechanistic insights relevant to other soft tissues studied in this thesis and highlighting the broader regenerative potential of vibration-based therapies.

Peri-implantitis is a progressive inflammatory disease that leads to bone loss, reduced bone density, and potential dental implant failure. As the condition worsens, the bone around the implant resorbs, causing implant instability and possible detachment. LIV has emerged as a promising non-invasive therapy, shown to boost bone density, enhance osteoblast activity, and slow or stop peri-implantitis progression. Studies and clinical cases suggest that LIV can improve bone or graft integration around compromised implants, offering a valuable adjunct treatment for implant-related bone loss (Horowitz and Kurtzman, 2023).

Regarding space travel, the absence of gravitational loading disrupts bone homeostasis, leading to significant bone loss during long missions. Traditional countermeasures like running and resistance training only partially slow this demineralization and are often uncomfortable and time-consuming for astronauts. Recent research suggests that brief daily exposure (under 20 minutes) to low-magnitude, high-frequency vibrations (15–35 Hz, <1 g) may help reduce or prevent space-induced bone loss, offering a simpler, cheaper, and more efficient alternative to current exercise regimens. Hawkey (2007) highlights the limitations of existing countermeasures and the potential of vibration therapy to better protect bone strength in microgravity.

In the same context, Touchstone et al. (2019) explored how LIV could counteract the negative effects of simulated microgravity (sMG) on mesenchymal stem cells (MSCs). They found that sMG reduced MSC proliferation and lowered key nuclear and mechanosensitive components (LaminA/C, Sun-2, Nesprin-2G) of the LINC complex, though without affecting cell viability. Notably, a twice-daily LIV regimen (90 Hz, 0.7 g) restored LINC component levels, nuclear stiffness, and partially rescued cell proliferation. This recovery depended on LINC integrity, as depletion of Sun-1 and Sun-2 blocked the effect. LIV also boosted YAP expression, a mechanotransduction factor suppressed by sMG. These results demonstrate LIV's potential to preserve stem cell function and nuclear integrity under mechanical unloading, with broader relevance for musculoskeletal health in microgravity, aging, and disuse conditions.

Pagnotti et al. (2016) showed in a mouse model of multiple myeloma, a disease marked by severe bone loss and marrow destruction, that LIV can reduce bone degradation and tumor progression. The authors used a mechanical loading protocol (90 Hz, 0.3 g, 15 minutes/day) for eight weeks on immunocompromised mice that had received injections of human myeloma cells. Animals treated with LIV showed less trabecular and cortical bone loss. According to histological analyses, the LIV group had significantly lower levels of marrow necrosis and a 31% decrease in tumor burden. Furthermore, as demonstrated by decreased serum TRACP5b levels and decreased eroded surface area in bone histomorphometry, LIV markedly inhibited osteoclast activity. Crucially, LIV did not promote tumor growth and might even have protective effects on the microenvironment of bone marrow, indicating that it might be a good non-pharmacological treatment option for cancer-induced bone disease.

The use of LIV as a non-invasive treatment strategy for osteosarcoma (OS), is investigated in the study by Xiong et al. (2024). In order to produce tumor-suppressive conditioned medium (CM) and induce tumor-suppressing cells (iTSCs), the researchers examine the effects of LIV on OS cells and mesenchymal stem cells (MSCs). The study identifies the ideal LIV conditions in the 10-100 Hz range and investigates the impacts on cell morphology, tumor progression, and cytokine expression using a combination of numerical simulations and experimental validation. The findings show that LIV not only reduces OS cell viability and motility but also encourages MSCs to differentiate into iTSCs. These findings propose LIV as an alternative biomechanic treatment.

A recent study by Xiong et al. (2024) investigates how LIV can be used therapeutically to slow the spread of cancer. It focuses on how LIV can influence the bone microenvironment and stimulate tumor-suppressive cells (iTSCs). In both breast and prostate cancer models, the authors show that LIV at 90 Hz and 0.7g can rewire lymphocytes and peripheral blood mononuclear cells (PBMCs) into iTSCs, which subsequently release a conditioned medium (CM) that inhibits tumor cell invasion, migration, and proliferation. The study also notes positive effects on bone remodeling, such as decreased osteoclast activity and increased osteoblast differentiation, underscoring LIV's dual function in addressing skeletal health and tumor burden. Notably, the authors note the translational potential of mechanical stimulation by reproducing some of these effects in viscous liquids using common lab equipment such as tube shakers. The intricacy of the in vitro models and the absence of in vivo validation in human tissues indicate that further research is necessary before therapeutic use, even though the molecular insights into pathways like LINC-SUN1 and c-Myc/K-Ras signaling are fascinating. Nevertheless, the study offers a solid basis for thinking about LIV as a non-invasive supplemental approach in the treatment of cancer and bone metastases.

3.2. Human Tissue Simulations and FEA

This section provides an overview of the current state-of-the-art in pertinent human tissue simulations.

For the purpose of examining the biomechanical reaction of the human thorax to low-frequency acoustic excitation, like that employed in high-frequency chest compression (HFCC) therapy, Uzundurukan et al. (2024) created a very thorough CT-based finite element

model (CT-FEM). The lungs, rib cage, trachea, and soft tissues were among the anatomically realistic structures included in their model. The material properties were allocated using the Voigt model for other thoracic components and Biot's poroviscoelastic theory for lung tissue. They determined that 28 Hz and 41 Hz are the primary thoracic resonance frequencies using frequency response function (FRF) and three-dimensional harmonic analysis, with 28 Hz being the most efficient in promoting lung tissue motion. The study showed that mechanical waves effectively travel across the thoracic cavity at certain resonance frequencies, especially increasing the lungs' displacement and energy density. A useful digital twin foundation for future studies on thoracic vibration is provided by these findings, which also support the use of vibroacoustic stimulation in airway clearance therapies and provide information pertinent to non-invasive mechanical therapies like low-intensity vibration.

Using ANSYS software, García-González et al. (2014) created a 3D finite element model of the human inner ear that is extremely detailed. One of the most anatomically comprehensive finite element models of the human inner ear was created and verified through their numerical investigation. To evaluate how vibrations spread throughout the auditory system, the authors used ANSYS to simulate both an isolated inner ear and a complete auditory system. According to their results, the displacement maxima along the basilar membrane shift predictably with frequency, moving from the apex at lower frequencies to the base at higher ones. This is consistent with experimental literature and supports the realism of the model. This study shows how finite element modeling may be used to investigate intricate, inaccessible biological systems like the inner ear and marks a major advancement in auditory biomechanics.

Using a finite element method to rebuild hyperelastic parameters, Mehrabian and Samani (2008) created a novel elastography technique to evaluate the mechanical behavior of breast tissue under massive deformations. They suggested modeling breast tissues as hyperelastic rather than assuming typical linear elasticity because they acknowledged that they show notable nonlinear strain-hardening responses during compression, particularly under circumstances like respiration or manual palpation. Two strain energy models, the Polynomial and Yeoh forms, were used in the study to recreate tumor-specific hyperelastic parameters and describe tissue function. Three different tissue types (adipose, fibroglandular, and tumor tissue) were incorporated into a simplified computational phantom that mimicked the geometry of the breast (a hemisphere attached to a cylinder). They estimated hyperelastic constants from displacement data by simulating 30% compression with ABAQUS and used an iterative inverse problem-solving technique. One of the key strengths of the study lies in its ability to rebuild precise tumor-specific material properties from exterior deformation data, even when beginning with initial estimates that were orders of magnitude off. Compared to the Polynomial model, the Yeoh model demonstrated faster convergence and better numerical stability, especially because it was less dependent on the strain invariant I_2 . By integrating three regularization techniques (truncated SVD, Tikhonov, and Wiener filtering) the authors also addressed the mathematical difficulties of resolving ill-conditioned inverse issues. Their findings feed the possibility that hyperelastic modeling and nonlinear elastography could enhance the contrast and diagnostic capacity of breast cancer imaging methods.

A thorough FEA of the human femur is presented by Kumar et al. (2024), with an emphasis on the dynamic behavior of the bone under both modal and harmonic loading conditions. The study reveals how bone shape and material properties affect vibrational

response by simulating natural frequencies and mode shapes under free-fixed and fixed-fixed boundary conditions using CT-based geometry. Through modal and harmonic analysis, resonance peaks and stress concentrations were found across a range of loads and acceleration inputs, with natural frequencies ranging from roughly 54 Hz to over 3,200 Hz. The study's main contribution is its illustration of how resonance close to natural frequencies can increase stress and deformation, which may raise the risk of fracture. This is particularly important in cases of trauma or repeated mechanical exposure. Despite the methodological robustness of the analysis, it simplifies real bone heterogeneity and may limit clinical translatability by assuming isotropic and homogeneous material behavior. However, this work serves as a useful reference for simulating bone dynamics under LIV loading.

Using non-linear, subject-specific parameters, Chen et al. (2013) created one of the most comprehensive and experimentally verified finite element models for examining dynamic breast motion. In addition to breast and thoracic structures, their model featured three separate skin layers with hyperelastic characteristics and realistic geometry acquired through high-resolution 3D photogrammetry. With a mean viscous damping ratio of 0.215 for a subject with an 80B cup size, the authors were able to capture the elastic and damping properties of breast tissue. Motion capture experiments involving walking, running, and stepping were used to validate the model, which demonstrated good agreement with actual breast displacement, especially in the vertical and anterior-posterior directions. Crucially, their modal analysis revealed natural frequencies in the 4–5 Hz range, which closely matched the step frequencies found in physical activity. Although the study lays a solid basis for modeling realistic breast dynamics, it is still mostly limited to applications involving external motion and bra design. By simulating the mechanical behavior of breast tissue under sinusoidal loading and concentrating on possible clinical applications, the current work expands on this.

Researchers have been using cutting-edge technologies more and more to enhance early diagnosis and treatment methods in response to the Mexican Social Security Institute's (IMSS) concerning statistics on breast cancer, which show that 28,000 new cases are diagnosed each year and 60% of them are at advanced stages. Of these, there has been a lot of interest in the use of ultrasound as a selective therapeutic technique. Using structural components such the cytoskeleton, actin filaments, microtubules, intermediate filaments, the nucleus, cytoplasm, and membrane, Gómez Girón et al. (2025) created a very detailed computational model of both healthy and malignant breast cells. The researchers discovered different resonance bands—27.1 to 31.27 kHz for MCF-7 cells and higher frequencies for MCF10A cells—by using modal analysis to determine the natural frequencies of malignant (MCF-7) and healthy (MCF10A) cells. Later harmonic analysis revealed that malignant cells responded with greater amplitudes at particular frequencies, indicating that they are more susceptible to mechanical stimulation. These findings provide a non-invasive and possibly affordable therapy option by laying the foundation for future frequency-targeted ultrasound-based therapies that may specifically damage cancer cells while preserving healthy tissue.

An extensive fluid-structure interaction (FSI) model of the human breast during lactation was created by Azarnoosh and Hassanipour in 2020. This model provides important information about the fluid dynamic and mechanical behavior of breast tissue under physiological suction. In order to simulate newborn suckling, their work applied a sinusoidal vacuum pressure at the nipple outlet and included a realistic multi-generation ductal network. In order to simulate interior tissue deformation and milk flow simultaneously, the study treated

breast tissue as a hyperelastic solid and milk as an incompressible Newtonian fluid. Visualizing how pressure waves travel through the ductal system and interact with deformable tissue to ultimately impact milk ejection was one of the work's major accomplishments. The significance of structural flexibility in efficient milk transfer was confirmed by this model, which also offered a framework for further study utilizing assistive or therapeutic technologies such as LIV.

LIV have a major effect on cellular behavior, especially in mechanically responsive tissues, according to recent studies. Uzer et al. investigated the effects of LIV on osteocyte communication and found that gap junction signaling between cells can be enhanced by even very little mechanical stimulation (0.15–1g, 30–100 Hz). Interestingly, the study showed that nuclear displacement was primarily driven by vibratory accelerations rather than fluid shear stress, indicating a direct biomechanical channel for mechanotransduction. This nuclear reaction was simulated using FEA, which demonstrated that vibratory loading causes more nucleus displacement than shear forces. The reasoning behind replicating vibrational inputs in soft tissues like the breast is supported by these findings, which set a solid precedent for the application of harmonic vibration in modeling and therapeutic contexts.

Recent developments in multiphysics modeling have highlighted how crucial it is to combine fluid dynamics and mechanical vibrations in order to reflect intricate real-world interactions. The study by Ouyang et al. (2022), which created a CFD-vibration linked model to forecast cavitation behavior in high-speed gear transmissions, is a noteworthy example. The work showed how dynamic structural excitation can greatly affect fluid pressure and vapor production in constrained geometries by combining CFD modeling of lubricant flow with finite element simulations of gear vibrations. Despite being used in mechanical systems, this technology is especially pertinent to biological settings like the nursing breast, where oscillatory motion caused by LIV may cause pressure changes and fluid displacement in ductal networks. A strong precedent for FSI techniques in soft tissue modeling is provided by the application of dynamic boundary conditions influenced by structural motion. Furthermore, using adaptive meshing and a two-phase cavitation model offers useful information about modeling phase changes or deformation-driven flow instabilities, which may be applied to the study of ductal collapse or milk flow under vibration. As a result, this work provides a conceptual link and methodological reference between fluid-mechanical behavior in biological systems and mechanical vibration investigations.

4. Design and Development

The setup, methods, assumptions and theoretical background used in the numerous experiments carried out throughout this investigation are thoroughly explained in this section. To guarantee clarity, reproducibility, and a good comprehension of the approach used, each experimental setting is documented.

4.1 Assumptions

Several key assumptions were taken into account to formulate the finite element simulations. They will be explained in detail in this subsection.

Firstly, the mechanical stimulus linked to LIV therapy was approximated in this study by applying a harmonic force of 1 Newton in all simulations. Delivering high-frequency, small-magnitude mechanical signals with minimal displacement amplitudes, usually below 1g acceleration, is what distinguishes LIV devices. A standardized 1 N input is a useful and physiologically reasonable approximation for simulation purposes, even though the actual forces applied in clinical devices may differ depending on frequency, surface area, and patient-specific circumstances. While staying within the mechanical bounds typically regarded as safe for biological tissues, this value enables consistent comparisons across tissues and frequencies. Furthermore, forces were defined in terms of its magnitude in Newtons rather than as a dimensionless amplitude, because for these simulations mode-superposition harmonic analysis was used.

For the purposes of this study, tissue homogeneity was assumed to improve computational efficiency. A simplified material definition is made possible by modeling biological tissues as homogeneous, which lowers simulation complexity while preserving the overall mechanical response to vibratory loading. Even though the density, stiffness, and composition of real biological tissues vary, this assumption allows for faster computation and more stable meshing, especially when performing multiple simulations or exploring wide frequency ranges. In preliminary analyses, this simplification offers a reasonable approximation for evaluating global deformation patterns and vibrational behavior, even though it might reduce the accuracy of local stress predictions.

Damping, the process by which mechanical energy in a vibrating system is gradually dissipated over time, was not incorporated into the simulations due to the lack of reliable and consistent data in the literature. Material-specific parameters, such as damping ratios and energy dissipation characteristics, are necessary for accurately modeling damping for biological tissues. These parameters vary greatly between studies and are frequently not well documented, particularly for soft tissues. Without definitive validation, adding these properties would significantly increase the models' level of uncertainty. Furthermore, the ANSYS Student license's restrictions on the use of sophisticated material models further limit the incorporation of damping. These factors led to the exclusion of damping in order to preserve model clarity and computational viability.

All tissues, including bone, were modeled as isotropic materials for this investigation. This simplified the definition of material behavior and decreased computational complexity by assuming that their mechanical properties were the same in all directions. The internal structure and directional loading patterns of biological tissues, especially bone, are known to produce anisotropic properties; however, the isotropic assumption makes simulations more manageable while still offering valuable information about overall stress distribution and deformation.

Based on the physiological presence of contractile myoepithelial cells surrounding the mammary ducts, the duct wall was modeled as muscle tissue in the CFD simulations. By contracting in response to oxytocin during lactation, these specialized cells play a crucial part in milk ejection. They actively propel milk through the ductal system because of their function, which is very similar to that of smooth muscle. As a result, using the mechanical characteristics

of muscle tissue to represent the duct wall provides a biologically accurate approximation that captures the duct's capacity to react dynamically to mechanical or hormonal stimuli.

The knee simulation used in this study purposefully omits cartilage and soft tissues in order to concentrate only on the joint's bony components. The goal of separating and examining the mechanical response of bone under LIV loading justifies this simplification. Modeling only bone structures offers a pertinent basis for initial analysis, since LIV may potentially affect subchondral bone remodeling and aid in delaying the progression of osteoarthritis in the knee. By focusing on bone-to-bone interactions, the model reduces complexity and preserves computational efficiency while capturing the dominant structural behavior. This method lays the groundwork for more intricate simulations in subsequent research while providing a useful initial approximation for comprehending how mechanical stimuli travel through joint bones, despite the absence of representation of cartilage mechanics.

4.2 Theoretical Background

4.2.1 Fundamentals Of Structural Dynamics

The general equation of motion governs the mechanical behavior of structures under dynamic loading (Kumar et al., 2024). The governing dynamic equilibrium equation for a linear system with multiple degrees of freedom is as follows:

$$M\ddot{x}(t) + C\dot{x}(t) + Kx(t) = F(t) \quad (1)$$

where:

- M is the mass matrix,
- C is the damping matrix,
- K is the stiffness matrix,
- $x(t)$ is the displacement vector,
- $F(t)$ is the external time-dependent force vector.

The equilibrium between externally applied forces, damping forces, inertial forces, and elastic restoring forces is described by this equation. Certain simplifications can be made to highlight particular dynamic behaviors, depending on the type of analysis.

4.2.2 Undamped Free Vibration (Modal Analysis)

The goal of modal analysis is to identify the system's inherent frequencies and mode shapes while ignoring damping effects and outside forces. Therefore, by setting $[C] = 0$ and $\{F(t)\} = 0$ in the general equation, the simplified form is as follows:

$$M\ddot{x}(t) + Kx(t) = 0 \quad (2)$$

Assuming a harmonic solution of the form:

$$x(t) = Xe^{i\omega t} \quad (3)$$

where X is the amplitude vector and ω is the circular natural frequency; substitution into the simplified equation yields:

$$(K - \omega^2 M)X = 0 \quad (4)$$

This is a classical eigenvalue problem, meaning that only for certain values of ω do non-trivial solutions for X exist. The system's natural frequencies and associated mode shapes, which describe its vibrational behavior in the absence of external excitation, are obtained by solving this equation.

4.2.3 Forced Vibration (Harmonic Analysis)

The goal of harmonic analysis is to ascertain the steady-state reaction when the structure is exposed to an external periodic force. In this instance, damping is disregarded for simplicity even though the entire dynamic equation is taken into account, which leads to:

$$M\ddot{x}(t) + Kx(t) = F(t) \quad (5)$$

The displacement vector is once more stated as follows, assuming a steady-state harmonic excitation and response:

$$x(t) = Xe^{i\omega t} \quad (3)$$

where the excitation frequency is denoted by ω . When this expression is entered into the dynamic equation, it produces:

$$(K - \omega^2 M)X = F \quad (6)$$

Determining how the structure deforms under cyclic loads at various frequencies is the aim of this frequency response problem. Harmonic analysis takes into account the impact of

an external dynamic load, in contrast to modal analysis, which solely looks at the system's intrinsic characteristics.

4.2.4 Fundamentals of Computational Fluid Dynamics (CFD)

A subfield of fluid mechanics known as computational fluid dynamics (CFD) solves and analyzes fluid flow issues using numerical techniques and algorithms. The basic conservation rules of physics, when applied to a control volume and written as partial differential equations, provide a mathematical description of fluid dynamics.

The Navier-Stokes equations, which express the conservation of mass (continuity), momentum, and energy in the fluid domain, are the principal equations controlling fluid flow.

4.2.4.1 Governing Equations

The conservation of mass (Continuity Equation) for an incompressible fluid is given by:

$$\nabla \cdot \mathbf{v} = 0 \quad (7)$$

where $\mathbf{v}$ is the velocity vector of the fluid.

The following is an expression for the conservation of momentum (Navier-Stokes Equations):

$$\rho \left(\frac{\partial \mathbf{v}}{\partial t} + (\mathbf{v} \cdot \nabla) \mathbf{v} \right) = -\nabla p + \mu \nabla^2 \mathbf{v} + \mathbf{f} \quad (8)$$

where:

ρ is the fluid density,

p is the pressure,

μ is the dynamic viscosity,

$\mathbf{f}$ represents body forces.

4.2.4.2 Application to Milk Flow Simulation

CFD is suggested as a tool in this thesis to simulate how milk flow behaves in the mammary ducts. Since milk is regarded as an incompressible Newtonian fluid, its viscosity is unaffected by shear rate. Laminar flow conditions can be anticipated since the flow is thought to take place inside small, confined ducts.

The fundamental CFD framework is still essential even though the dynamic interaction between LIV and milk flow was not fully addressed in this work due to later explained limitations. Either way, the results presented in further sections serve as a foundation for

upcoming modeling studies that examine the impact of external mechanical stimuli on breastfeeding dynamics by coupling fluid flow with harmonic mechanical loads.

4.3 Breast and Lactation

These simulations focused on the mechanical response of breast tissue to harmonic loading. These tests were conducted utilizing a combination of mode-superposition harmonic analysis and modal analysis to capture dynamic behavior efficiently without requiring too much processing power. By analyzing the resulting stress and strain distributions, it is possible to identify specific breast regions that may be exposed to stronger mechanical stimulation; these locations may be linked to better blood circulation or lymphatic drainage.

The passive flow of milk inside the ductal system was also examined independently to establish a physiological baseline.

4.3.1 Adipose Tissue Cube

As an approximation to initiate simulations, it was decided that the first logical step was to create a 3D cube composed of a material that has a great presence in the human breast. It was decided that adipose tissue could be used due to its large presence in the human breast and easier accessibility than pure glandular tissue, whose properties were not as easily found in the current literature.

For this reason, fatty tissue was represented by a basic three-dimensional model as a first approximation. ANSYS SpaceClaim was used to create a simple cube shape that measured $30 \times 30 \times 30$ mm. With a density of 950 kg/m^3 , obtained from a study from Martin et al. (1994), a Young's modulus of 1 KPa (Comley and Fleck, 2010), and a Poisson's ratio of 0.49 (Fougeron et al., 2024), the material properties were determined under the assumption of isotropic elastic behavior.

After the geometry was automatically meshed, 8,281 elements and 1,728 nodes were produced. To mimic attachment to an immobile base, a fixed support was placed on the cube's bottom face. The first 7 modes were extracted. The harmonic analysis was driven by the results of the modal analysis. To simulate external mechanical stimulation, a 1 N force was applied to the cube's top face in the X-direction during the harmonic study. The frequency sweep of the harmonic analysis was set to a range between 0 and 30 Hz.

4.3.2 Breast-like Geometry of Adipose Tissue

The purpose of this second simulation was to make the geometry more realistic. It was created using ANSYS SpaceClaim. As seen in Figure 2, the breast shape was simplified combining a cylinder and a hemisphere. To guarantee consistency across simulations, the material was defined in the Engineering Data section of ANSYS using the same material properties used in the fat tissue cube simulation.

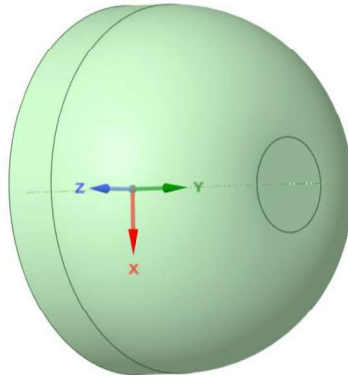


Figure 2. Geometry resembling a human breast, created using SpaceClaim.

The mesh was automatically created by ANSYS Mechanical, generating 2,532 nodes and 1,361 elements. It is displayed in Figure 3.

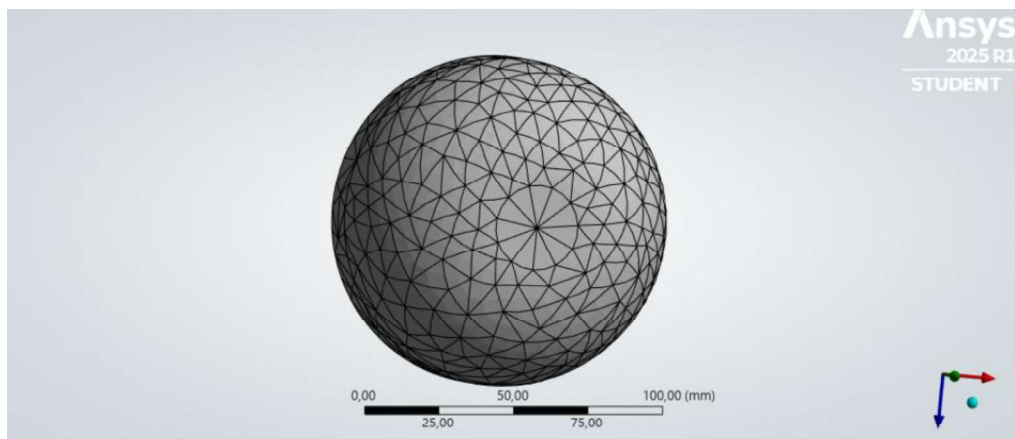


Figure 3. Breast-like geometry meshed.

A fixed support was introduced in the posterior face of the geometry, resembling the attachment to the thoracic wall.

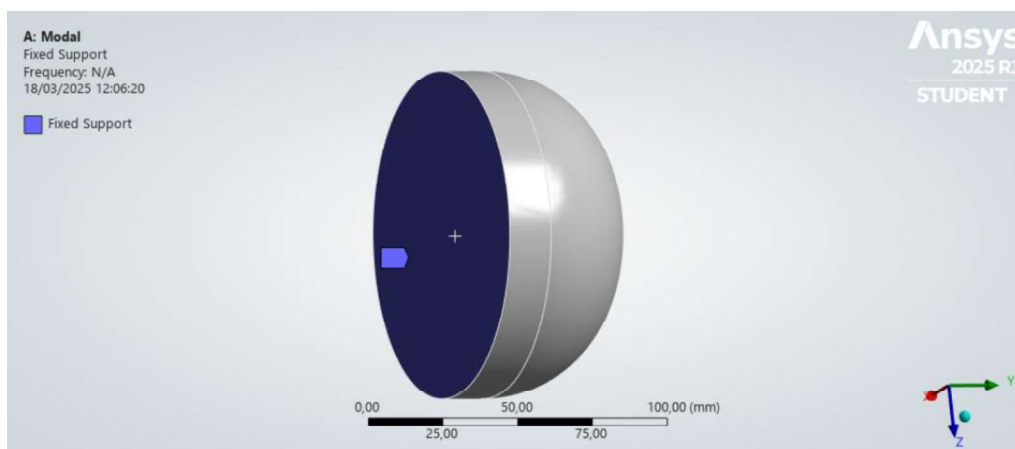


Figure 4. Fixed support introduced in the breast-like geometry.

Following the introduction of the fixed support, a force of 1 N was applied to the areola section of the breast-like geometry.

4.3.3 Breast-like Geometry of Breast Tissue

After first using fat tissue to approximate soft tissue response, the next set of simulations was inspired by the work of Chen et al. (2013), who developed a detailed finite element model of the human breast. Their model incorporated complex anatomical layers, including the thoracic wall, breast tissue, and multiple skin layers, offering high biological accuracy. In this study, however, software and computational limitations (particularly within the ANSYS Student framework) made it difficult to replicate that level of detail. As a result, the same simplified breast-like geometry used in the previous section was applied here.

Given the mechanical characteristics of biological soft tissues like the breast, hyperelastic material models were initially considered ideal for the simulation. Hyperelastic materials are capable of withstanding large, reversible deformations—up to 40% strain—without exhibiting plastic behavior. Their nonlinear stress-strain response cannot be accurately captured using classical linear elastic models governed by Hooke's Law (Melly et al., 2021). Among the available hyperelastic formulations, the neo-Hookean model is widely adopted in breast biomechanics due to its simplicity and reasonable approximation of soft tissue behavior. However, due to the limitations of the ANSYS Student license, neo-Hookean modeling could not be used; materials then needed to be manually created using isotropic elasticity and density.

Reported values for the equivalent Young's modulus of breast tissue vary considerably, ranging from 0.2 to 60 kPa. This variability can be attributed to both inter-individual differences in tissue composition (Lorenzen et al., 2003) and methodological differences across studies. As highlighted by Mîra et al. (2018), the mechanical characterization of breast tissue generally falls into two methodological categories: *ex vivo* indentation tests, which provide direct mechanical measurements, and *in vivo* imaging techniques combined with finite element modeling, allowing non-invasive property estimation.

In spite of these issues, a novel material labeled "Breast" was created within the ANSYS Workbench environment. In line with the approach taken by Chen et al. (2013), the breast tissue was modeled as an incompressible, homogeneous, and isotropic solid. The parameters introduced were those measured in an *in vivo* setting in the same study. According to Chen et al. (2013) and Katch et al. (1980), a density of 1000 kg/m³, or that of water, was chosen, and its elastic behavior was defined by a Poisson's ratio of 0.495 and a shear modulus of 0.34 kPa.

In order to ensure precise and reliable simulation results, suitable boundary conditions and mesh settings were established within the ANSYS environment to prepare the breast geometry for modal and harmonic analysis. The geometry was subjected to an automatic structured mesh, yielding 21,534 nodes and 12,283 elements overall. A fixed support boundary condition was applied to the model's posterior surface to replicate the mechanical attachment to the thorax and the anatomical constraint of the breast being anchored to the chest wall. The first 1000 vibration modes were selected to be extracted for the modal analysis. This large number of modes was selected to try to cover the entire frequency spectrum

pertinent to LIV devices. A range of different frequencies were utilized for the harmonic analysis.

4.3.4 Two-Breast Realistic Geometry

A realistic breast geometry created by Foad S. Farimani was retrieved from GrabCAD as an IGS file. It was imported directly into ANSYS DesignModeler. The scale was off, with a measure of 0.29 cm across the chest. Because the average adult female thorax width is around 26 cm in females aged 20-39 years old (Obikili and Okesina, 2006), scaling needed to be performed. Using the scale transformation from DesignModeler with a scaling factor equal to 90, the geometry was successfully scaled. With the original format, meshing could not be successfully performed. For that reason, preprocessing was carried out. SpaceClaim's Shrinkwrap tool was used as cleanup attempts didn't work. This function produces a waterproof outer shell, which is ideal for FEA. It was then converted into a meshable solid. In Figure 5, the processed geometry may be observed.



Figure 5. Two-Breast Geometry.

The same material as in the previous section, "Breast", was created. Like Chen et al. (2013), the breast tissue was deemed incompressible, homogeneous, and isotropic. According to Chen et al. (2013) and Katch et al. (1980), a density of 1000 kg/m^3 , was chosen. Its Poisson's ratio was 0.495 and its shear modulus 0.34 kPa.

To prepare the two-breast geometry for modal and harmonic analysis, the mesh was automatically generated, yielding 50,110 nodes and 28,701 elements. A fixed support boundary condition was applied to the model's posterior surface to replicate the thorax. The first 1000 vibration modes were extracted in the modal analysis, to try to cover a decent frequency range compatible with LIV devices. 1 N forces were applied above the areola area. Different frequencies were specified during harmonic loading.

4.3.5 CFD Passive Milk Flow

To study the distribution of milk through the ductal system, a steady-state CFD simulation was conducted. The setup described in the following section mimics passive milk flow without mechanical stimulation.

In order to simulate a human milk duct, ANSYS DesignModeler was used to create a simple cylinder geometry. The cylinder's diameter and length were determined to be 2 mm

and 10 mm, respectively. The Ramsay et al. (2005) study, which used ultrasound imaging to examine the anatomy of the nursing breast, served as the basis for the selection of these values. There, it was discovered that the left and right breasts' mean duct diameters were 1.9 ± 0.6 mm and 2.1 ± 0.7 mm, respectively. The same study was used to determine the cylinder length value. It revealed that the first branch of the main duct was 8.20 ± 6.27 mm from the nipple base in the left breast and 7.00 ± 3.98 mm in the right.

The geometry was automatically meshed using ANSYS Mechanical and can be observed in Figure 6. An element size of 0.1 mm was utilized. This configuration resulted in 86,904 nodes and 77,366 elements.

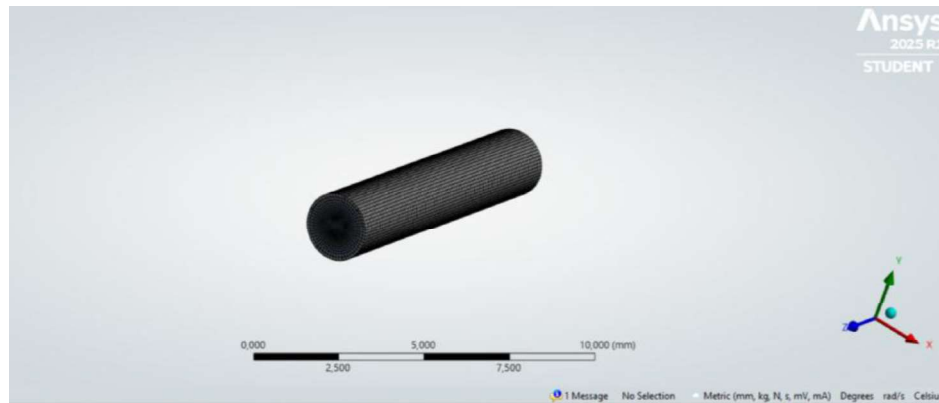


Figure 6. Meshed ductal-like geometry.

Once meshed, the setup of the model took place. The parameters chosen in the ANSYS Fluent launcher were 3D, Double Precision and a Pressure-Based solver. The model was set to laminar, since the flow is slow and non-turbulent.

For the fluid portion of the mesh, a new material called Milk was developed. Due to the fact that human milk was considered as an incompressible Newtonian fluid in previous studies (Almeida et al., 2011) with an approximate density value of 1030 kg/m^3 (Negin Mortazavi et al., 2017), this was the value entered in ANSYS Fluent to run the simulation. Regarding viscosity, Negin Mortazavi et al. (2017) highlight that the value of human milk's viscosity can vary significantly. It's indicated that it can range from $50 \times 10^{-6} \text{ m}^2/\text{s}$ on the first day postpartum to $1.66 \times 10^{-6} \text{ m}^2/\text{s}$ as lactation progresses. When converted to the units used by ANSYS Fluent, these values correspond to a viscosity range of 0.0017 to 0.0517 $\text{kg}/(\text{m} \cdot \text{s})$. For this simulation, an arbitrary, more leaning towards the middle to late lactation period, value of 0.003 $\text{kg}/(\text{m} \cdot \text{s})$ was chosen.

To the best of the author's knowledge, the current literature does not contain the precise density value of the lactating duct wall, which is necessary to simulate the solid part of the CFD simulation. However, the decision was made to model the duct wall as though it was composed of striated muscle because of the composition of the mammary ducts, which are surrounded by a discontinuous layer of contractile myoepithelial cells and lined by a luminal layer of either columnar or cuboidal epithelial cells (Owens et al., 2013). The National Institute of Standards and Technology's (NIST) Material Properties Database for biological tissues provided the striated muscle's density value (NIST, 2025). 1040 kg/m^3 is the equivalent of this value.

A velocity inlet condition was applied at the inlet surface with a consistent velocity magnitude of 0.05 m/s. In order to guarantee laminar flow and replicate baseline conditions free from the effects of external stimulation or suction, this value was selected. With a gauge pressure of 0 Pa, a pressure outlet boundary condition was applied at the outlet surface, reflecting atmospheric pressure and enabling the fluid to freely exit the domain.

4.3.6 CFD Passive Flow Realistic Geometry

To produce a more realistic geometry, GrabCAD was browsed. A carotid bifurcation geometry, created by an user named Aaron, was selected to mimic a bifurcation of the mammary ducts. After measuring the geometry (which has the measure of an artery), it was scaled down using a scaling factor of 0.333, resulting in a duct diameter of approximately 2 mm (as per measures of Ramsay et al., 2005). The meshing was carried out (26,230 nodes and 133,423 elements) and the inlet and outlet surfaces were set using named selection. It was introduced into ANSYS Fluent, where inlet and outlet conditions are observable.

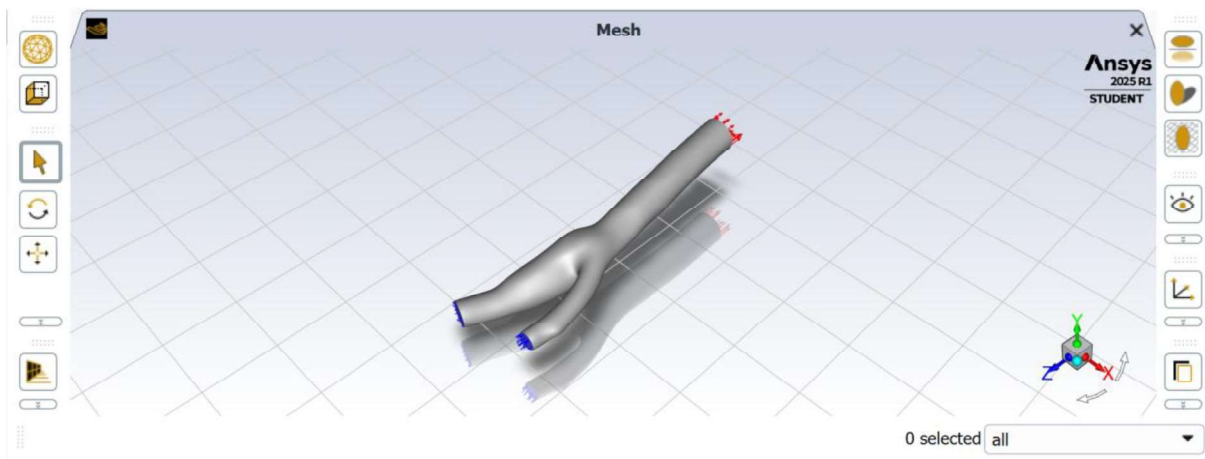


Figure 7. Realistic duct geometry with inlet and outlet boundary conditions.

The model was set up as laminar in ANSYS Fluent, and the gravity value was added. Negin Mortazavi et al. (2017) provided the density of human milk, which was determined to be 1030 kg/m³, as well as the viscosity value, which was 0.003 kg/(m·s). The outlet was set up as pressure-based, and the velocity at the inlets was set at 0.05 m/s. To get the simulation results, these parameters were set, the solution was initialized, and the calculation was performed 150 times.

4.4 Bone

4.4.1 Femur

Modal analysis is very useful for the identification of vibrational characteristics, encompassing mode shapes and natural frequencies of the human femur (Kumar et al., 2024).

4.4.1.1 Simplified Geometry

For computational efficiency, bone is typically assumed to have isotropic material properties (Geraldes and Phillips, 2014). For this same reason, the simulations will be acquired utilizing isotropic elasticity, with known parameters extracted from Matweb. The elasticity was derived from the shear modulus and the Poisson ratio, with values equal to 5.6 GPa and 0.24, respectively. The density was obtained from Matweb too, with a value of 2060 kg/m³.

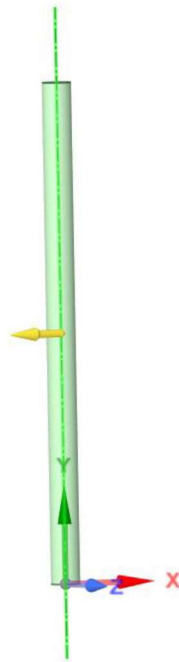


Figure 8. Femur-like geometry.

The femur bone was simplified utilizing a cylinder, which was created using SpaceClaim. The length of the cylinder was determined to be 40.24 cm (Vasilopoulos et al., 2020), with a diameter of 2.34 cm (Arun and Jadhav, 2016). The mesh was automatically generated, rendering a total of 2,594 nodes and 484 elements.

Two fixed supports were applied, one in each of the circumferences which would be the femur heads. Both the distal end and the femoral head are prohibited from moving in this fixed-fixed boundary condition. The first 25 modes were extracted in hopes of covering the range of the LIV device.

The outcomes of the previous modal analysis served as the basis for setting up the harmonic response analysis. To guarantee uniformity in boundary conditions, the two fixed supports specified in the modal setup were kept. To capture potential resonant effects and cover the entire operational bandwidth of a LIV device, a frequency range of 30 Hz to 10,200 Hz was specified.

The mid-region of the femur's anterior surface, which is where the LIV device is expected to make contact, received a harmonic force of 1 N. In accordance with the principal

direction of applied loading, the force was applied along the X-axis. Notably, because the harmonic analysis was done using the mode-superposition method, the force was defined in terms of its magnitude in Newtons rather than as a dimensionless amplitude.

4.4.1.2 Realistic Geometry

After a first approximation, a more complex geometry was tried. The geometry used in this simulation was retrieved from the NIH 3D Print Exchange database by searching for the keyword “knee”, as this anatomical region will also be referenced in subsequent sections of the study. The downloaded STL file was imported into ANSYS SpaceClaim, where it was first separated into its individual facets. The Autofix tool was applied to identify and resolve any minor geometry issues. From the resulting components, only the femur facet was selected for analysis, while the remaining elements were discarded. Once isolated, the femur facet was converted into a solid body.

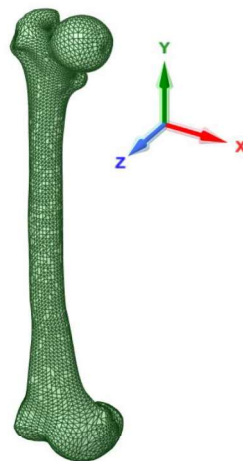


Figure 9. Isolated femur geometry, created in SpaceClaim.

Assuming isotropic behavior, the material properties were defined using the same methodology as the last model. The density was set at 2090 kg/m^3 , and the elastic properties were calculated using the given shear modulus and Poisson's ratio. With 47,019 nodes and 26,636 elements, a fine mesh was created to guarantee adequate resolution for precise simulation results. In order to replicate anatomical constraints, boundary conditions were created by applying two fixed supports at the proximal and distal ends of the femur. Same modes, harmonic ranges and forces were applied in this simulation.

4.5 Knee

By using the keyword "knee" to search the NIH 3D Print Exchange database, the geometry used in this simulation was found. For additional processing, the relevant STL file was obtained and downloaded. Figure 10 displays the final model.

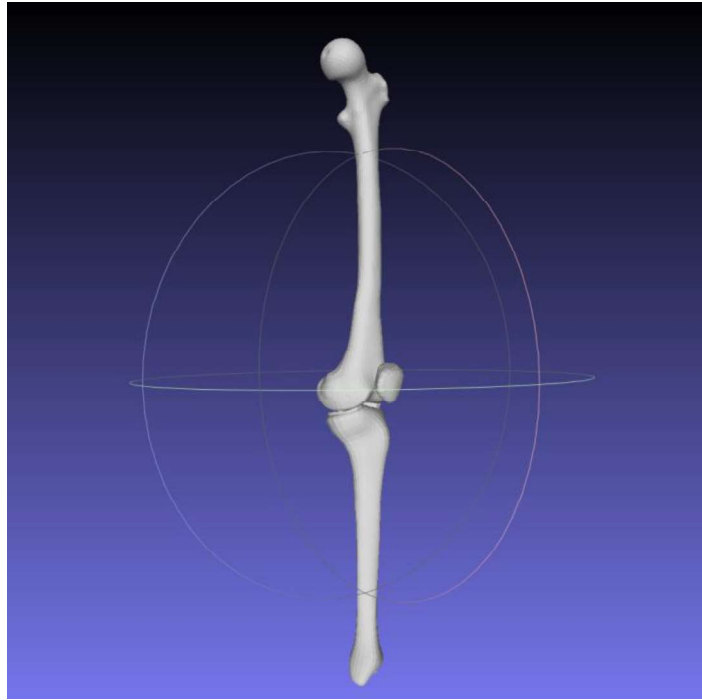


Figure 10. 3D Knee Geometry, retrieved from National Institutes of Health, n.d.

The model needed meticulous preprocessing in ANSYS SpaceClaim because of the geometry's intricacy. Separating every facet of the imported STL file was the first step. The Autofix tool was then used to process these facets. The facets were transformed into solid bodies after verification. To maintain continuity, the Combine function was only used to combine the solids that were in physical contact. Since ANSYS Mechanical cannot mesh STL files directly without first turning them into solid geometry, this preprocessing step is crucial. Errors or failures in the meshing process would result from this conversion. For dependable finite element meshing and subsequent simulation, turning the STL into a solid model is therefore essential.

The same material properties used in the femur simulation were specified in ANSYS Workbench. The meshing was automatically carried out by ANSYS Mechanical, resulting in a fine mesh with 78,555 nodes and 45,138 elements. Two fixed supports were introduced in both extremes. Six modes were extracted. The modal analysis results were streamlined into the harmonic analysis pipeline, where a 1 N force was introduced in the contact region.

4.6 Ovaries

The geometry used was retrieved from the NIH 3D Print Exchange database as a STL file and can be observed in Figure 11.

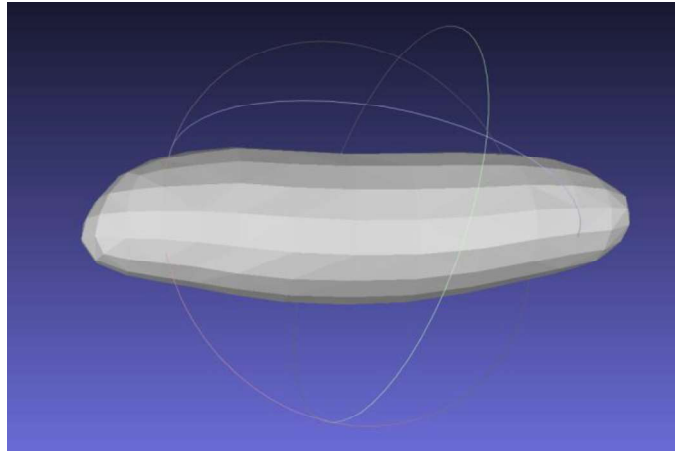


Figure 11. STL file of a human ovary.

Values from Hopkins et al. (2021) were used to define the material properties of the ovarian tissue. According to their research, the ovary is a comparatively soft tissue with mechanical qualities similar to those of the kidney or adipose tissue. In particular, the organ's compliant nature was demonstrated by the reported mean Young's modulus of 3.3 ± 2.5 kPa. In keeping with the presumptions of the initial research, a Poisson's ratio of 0.45 was also used in this investigation. The ovary's response to vibratory loading conditions was modeled using these parameters. The density value was retrieved from Noblet et al. (2018) and was equal to 1.05 g/cm^3 .

The meshing was carried out automatically using ANSYS Mechanical, yielding a total of 6,056 nodes and 3,409 elements. Two fixed supports were added in the extremities of the geometry, using face selection. They can be observed in Figure 12.

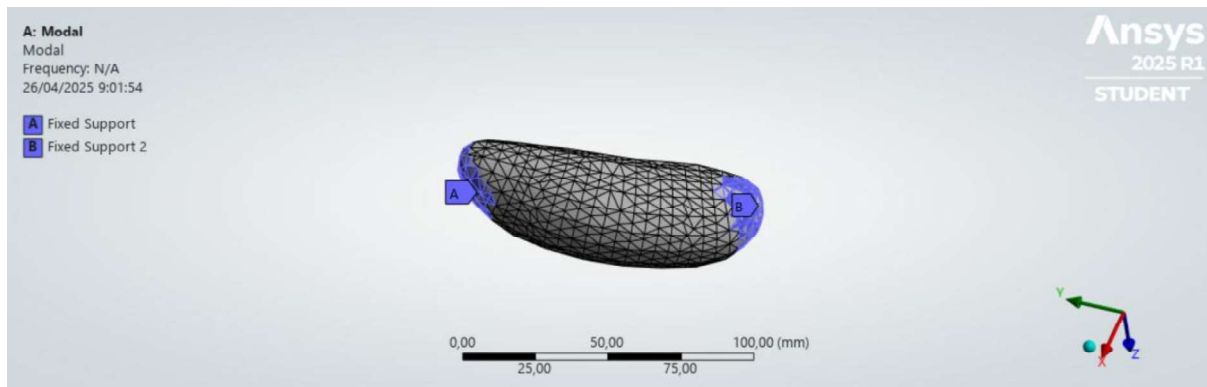


Figure 12. Ovary geometry with two fixed supports.

For mode extraction, first it was done using 25 modes and later 1000 modes to try to cover resonance in a large range. For harmonic analysis, the results from the modal analysis were streamlined. The studied frequency range was equal to 30-1000 Hz and the applied force was 1 N.

5. Experiments And Results

5.1 Breast And Lactation

5.1.1 Adipose Tissue Cube.

5.1.1.1 Modal Analysis

Around 3.6712 Hz was the first natural frequency to be extracted. From the maximum deformation at the top surface to the minimum deformation at the fixed bottom surface, the deformation distribution exhibits a gradual gradient. At the top face of the cube, the maximum total deformation measured was roughly 361.89 mm, whereas at the fixed support, the minimum deformation was 0 mm.

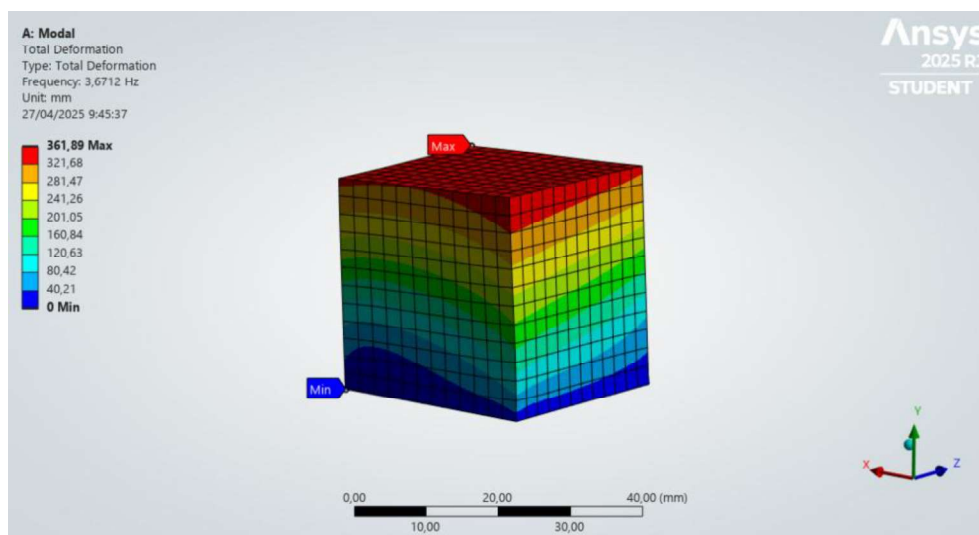


Figure 13. Total deformation in the first mode of an adipose tissue cube.

The first seven modes and the 56th one are presented in Table 1 below. The 56th mode is important to understand further results.

Mode	Frequency (Hz)
1	3.6712
2	3.6712
3	4.6349
4	8.9899
5	9.3086
6	9.3086
7	11.262

56	30.001
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Table 1. Modes of the adipose tissue cube.

5.1.1.2 Harmonic Analysis

The red and yellow zones in the fat tissue cube's harmonic response analysis at 30 Hz reveal a severely deformed shape with notable localized areas of stress concentration. Approximately 0.00839 MPa, or 8.39 kPa, is the highest principal stress measured, and approximately -0.00371 MPa, or -3.71 kPa, is the lowest. The resonance effects leading to these results will be discussed in further sections.

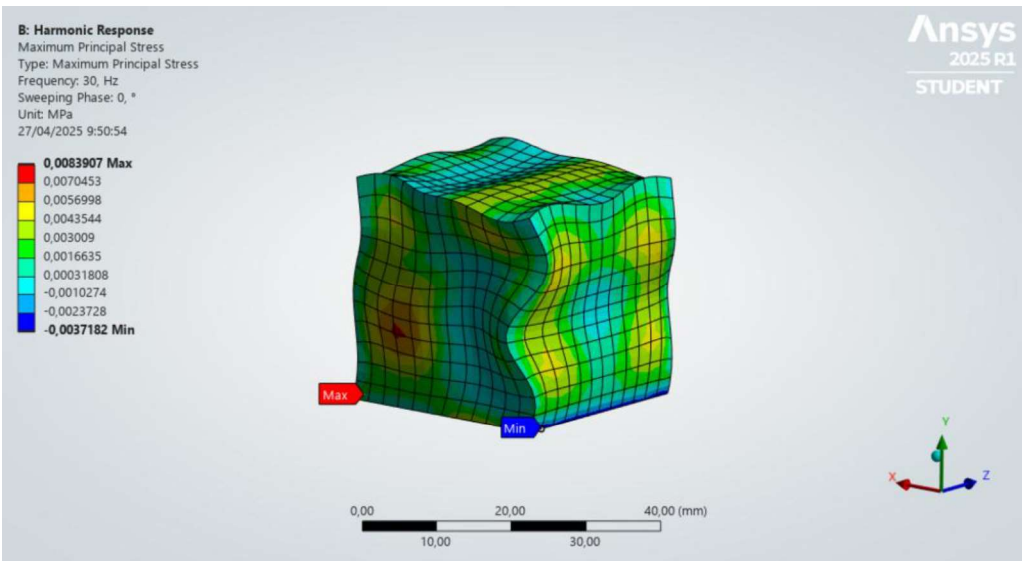


Figure 14. Maximum principal stress of an adipose tissue cube at 30 Hz.

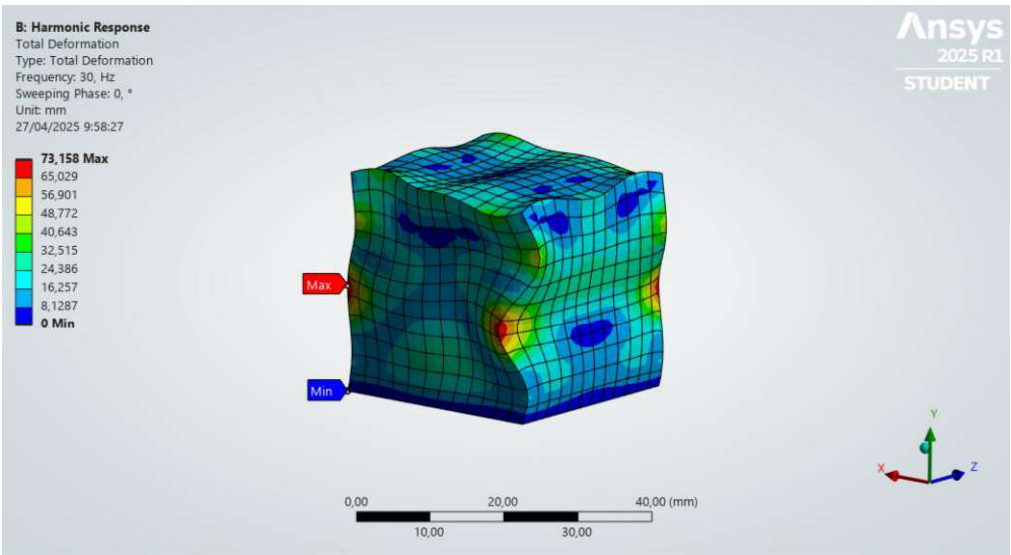


Figure 15. Total deformation of an adipose tissue cube at 30 Hz.

The harmonic response at 30 Hz shows that the maximum total deformation measured was approximately 73.158 mm, concentrated in an area near one of the cube's side faces, as

indicated by the red zone in Figure 15. The deformation field exhibits a continuous gradient from the maximum deformation area toward the lower displacement regions shown in blue. The minimum total deformation of 0 mm was used to represent the fixed support boundary condition at the base of the cube.

5.1.2 Breast-like Geometry of Adipose Tissue

5.1.2.1 Modal Analysis

The first extracted natural frequency, according to the modal analysis, was roughly 2.864 Hz.

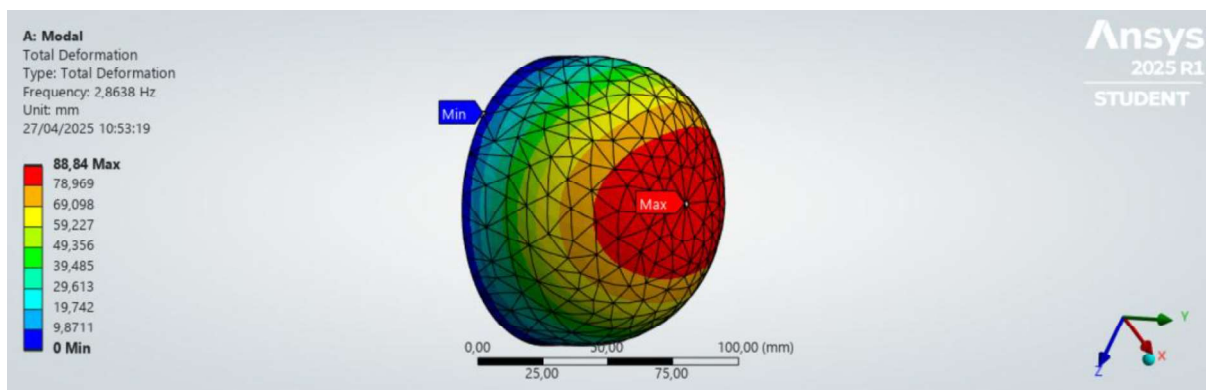


Figure 16. Total deformation corresponding to the first mode shape of the breast-like adipose tissue model.

5.1.2.2 Harmonic Analysis

As seen in Figure 17, the maximum principal stress measured was 0.43675 kPa and was concentrated close to the model's side region. With the lowest principal stress recorded at roughly -0.11499 kPa, the stress field diminishes to the outskirts from the maximum region.

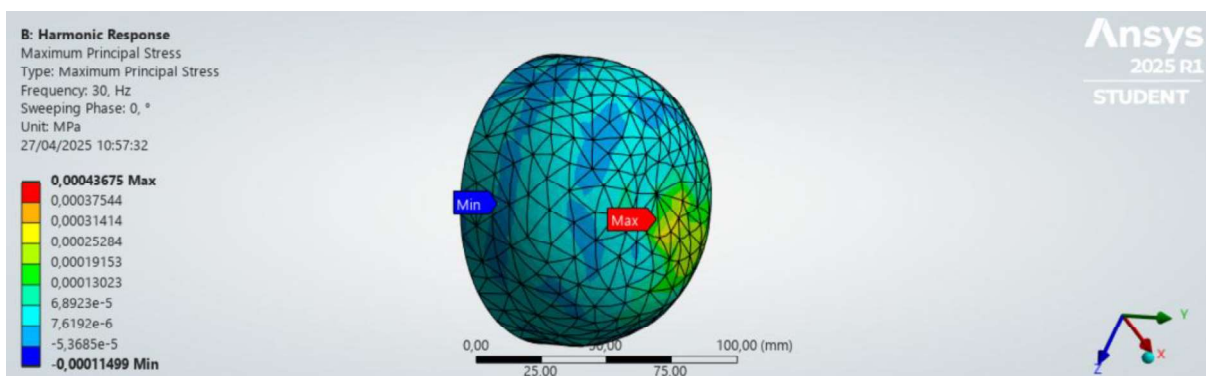


Figure 17. Maximum principal stress of the breast-like adipose tissue model at 30 Hz.

Regarding total deformation, Figure 18 below indicates the front region of the model, where the maximum total deformation measured was 10.467 mm. At the rest of the geometry, a minimum of 0 mm can be observed.

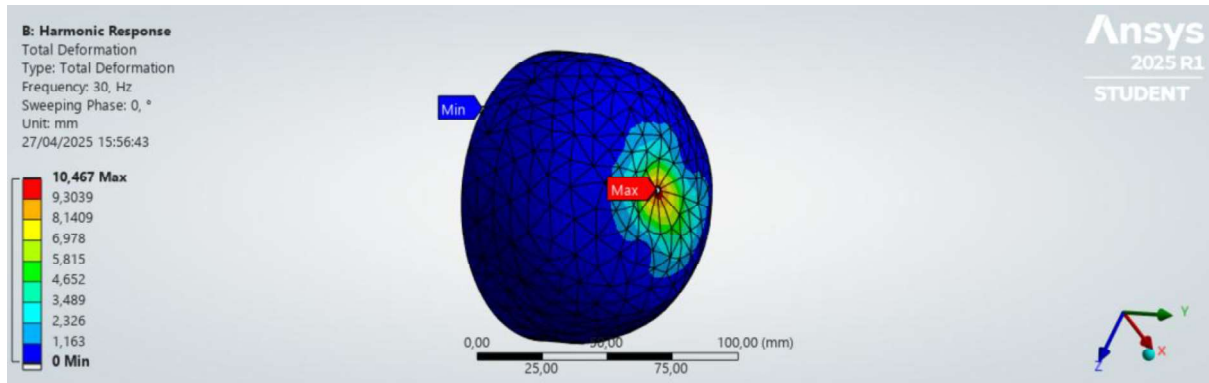


Figure 18. Total deformation of the breast-like adipose tissue model at 30 Hz.

5.1.3 Breast-like Geometry Made of Breast Tissue

5.1.3.1 Modal Analysis

In this section, the results of the modal analysis conducted on a simplified 3D breast model are presented. A total of 1000 modes were extracted, reaching a maximum frequency of 435.84 Hz, as observed in Figure 19. The first six modes are presented in Table X below.

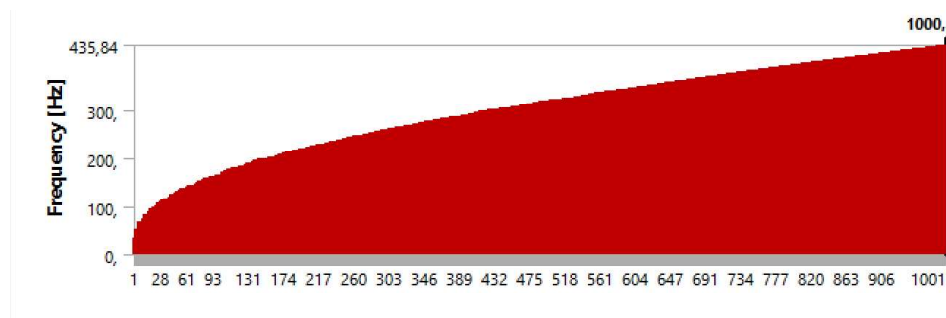


Figure 19. Plot displaying the number of modes and the frequency achieved (in Hz).

Mode	Frequency (Hz)
1	27.919
2	27.933
3	34.901
4	51.787
5	54.234
6	54.243

Table 2. First six modes of the breast tissue model.

About 27.919 Hz was the first natural frequency to be extracted (Figure 20). The maximum displacement was 86.305 mm, centered on the model's front face.

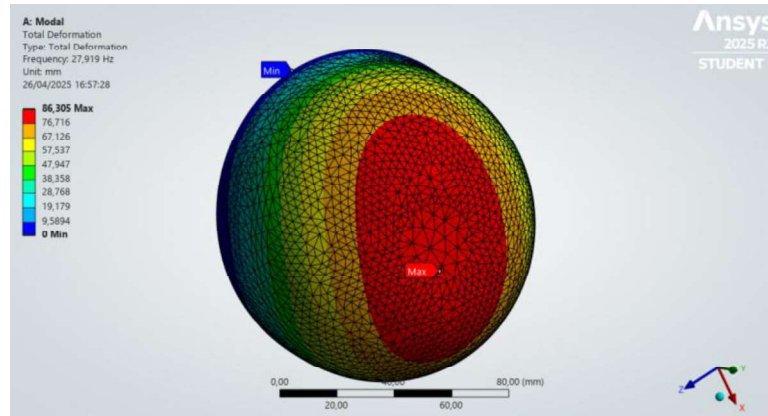


Figure 20. Total deformation of the first mode.

5.1.3.1 Harmonic Analysis

Although no additional modes were extracted beyond 437 Hz, the harmonic analysis was deliberately extended to cover a broader frequency range, from 20 Hz up to 1,000 Hz, to account for the potential operating bandwidth of the LIV device. The goal wasn't just to capture resonance phenomena but also to assess the tissue's general vibrational response and mechanical stimulation across clinically relevant frequencies. Even without natural resonances above 437 Hz, applying higher-frequency excitations can still produce meaningful forced responses in the tissue, making it reasonable to extend the harmonic sweep beyond the last extracted mode.

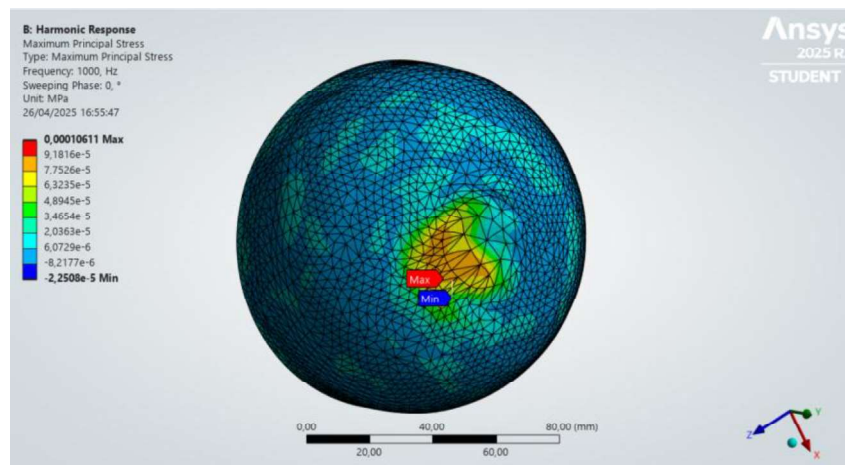


Figure 21. Maximum principal stress of a breast-like model at 1,000 Hz.

The highest stress value, approximately 0.10611 kPa, was concentrated in a small area near the model's nipple, as shown by the red zone in Figure 21. The stress gradually decreases outward from this area, reaching its lowest principal stress at roughly -0.02251 kPa.

A second experiment was carried out, now defining the frequency range between 20 and 10,000 Hz, in order to see the behavior of the tissue at these higher frequencies.

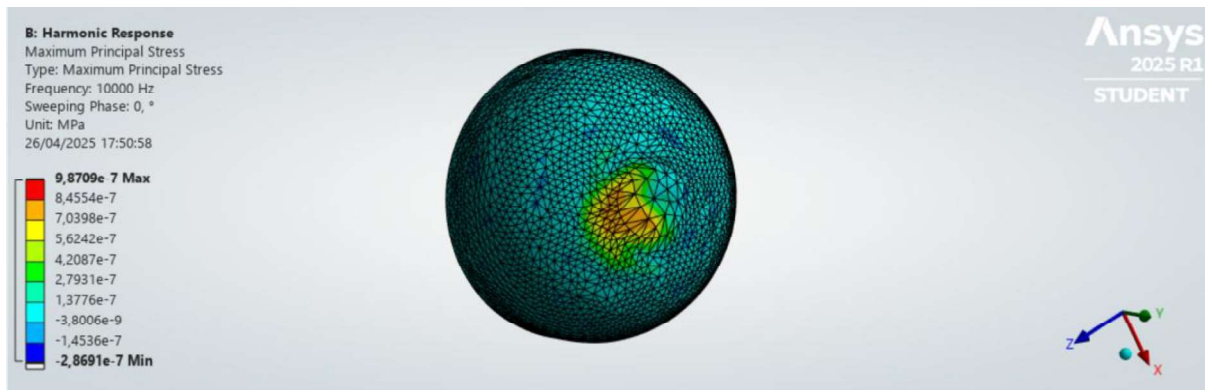


Figure 22. Maximum principal stress of a breast-like model at 10,000 Hz.

Approximately 0.00098709 kPa was the highest stress, which is really low despite that high of a frequency. The stress distribution remains relatively smooth, with a gradual decrease in magnitude away from the peak stress zone.

The third experiment was conducted with a frequency range extending from 20 Hz to 40 Hz. The stress response can be observed in Figure 23.

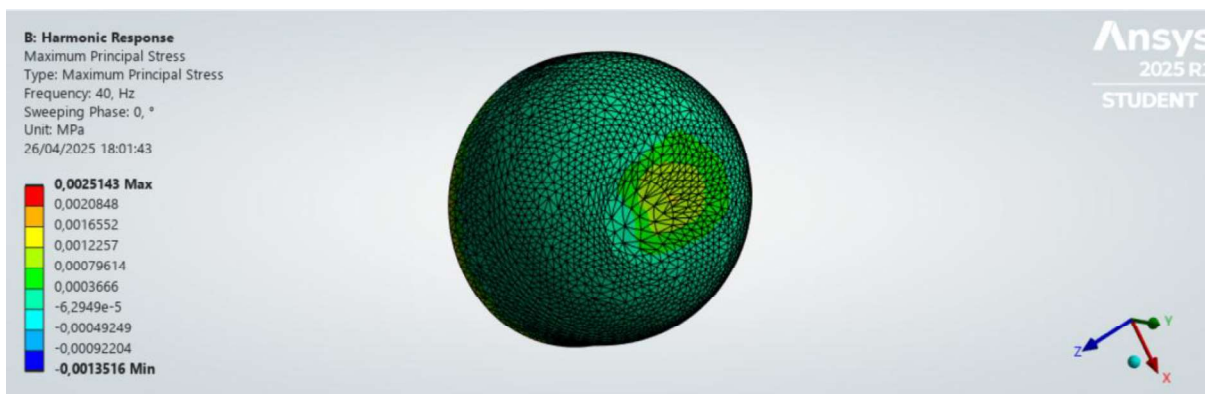


Figure 23. Maximum principal stress of a breast-like model at 40 Hz.

The red zone indicates that the highest measured principal stress was approximately 2.5143 kPa, which is way higher than at 1,000 and 10,000 Hz.

5.1.4 Realistic Two-Breast Model

5.1.4.1 Modal Analysis

The findings of a modal analysis conducted on the two-breast model are shown in Figure 24. Around 1.5162 Hz is when the first mode shape appears. With 1000 modes extracted, a frequency of only 18.578 Hz was achieved. Since soft biological tissues like fat, glandular tissue, and skin normally have elastic moduli in the kilopascal range, these low modes are primarily caused by their naturally low stiffness. Consequently, even when many modes are examined, these materials inherently display low-frequency deformation modes.

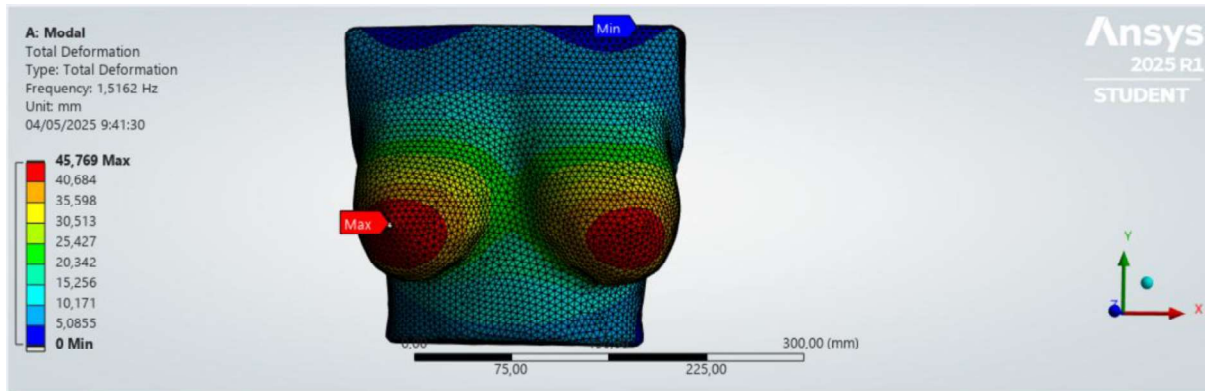


Figure 24. First mode of the two-breast model at 1.5162 Hz.

5.1.4.1 Harmonic Analysis

The maximum principal stress distribution under a harmonic excitation at 20 Hz is shown in Figure 25. The stress reaches its maximum at about 0.000374 MPa, which is a low value that is consistent with LIV devices.

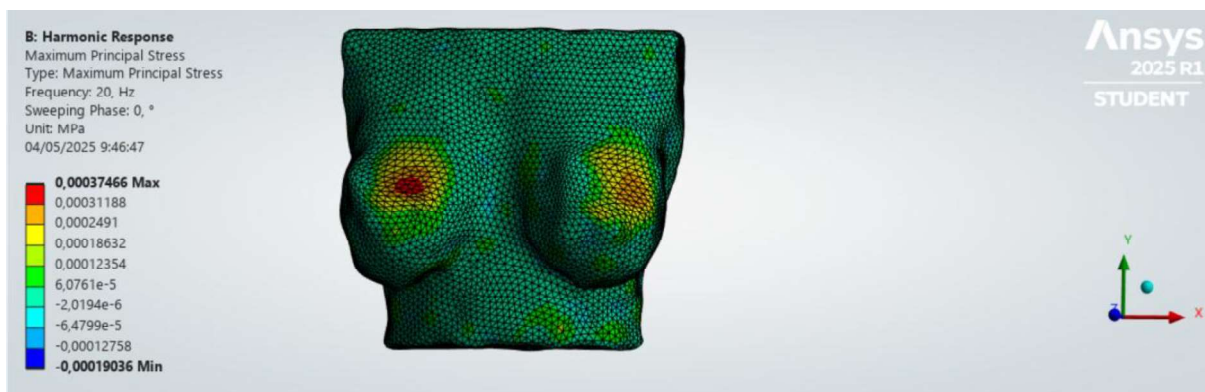


Figure 25. Maximum principal stress of the two-breast model at 20 Hz.

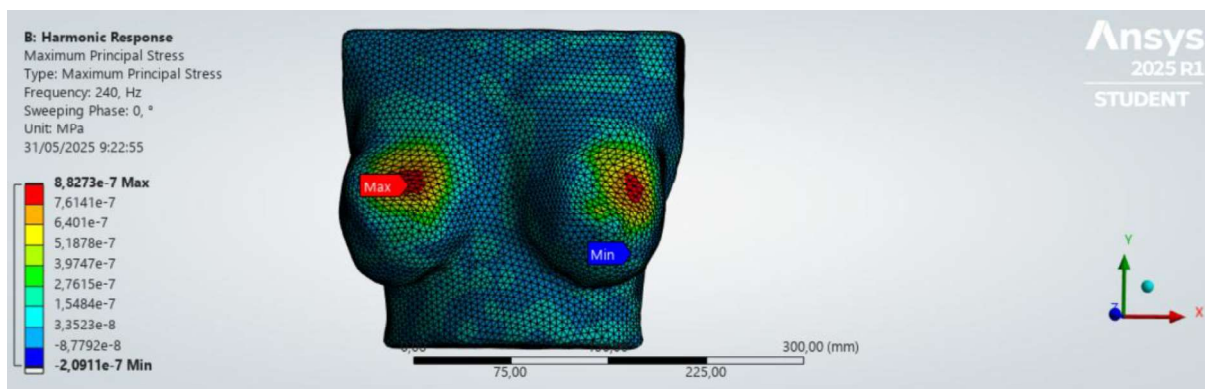


Figure 26. Maximum principal stress of the two-breast model at 240 Hz.

When analyzed at 240 Hz (Figure 26), the stress reaches its maximum at about 8.83e-7 MPa, which is lower than at 20 Hz.

In Figure 27 we see the stress distribution of the two-breast model at 1000 Hz. The maximum stress is equal to 5.07×10^{-8} , which is again lower than both 20 Hz and 240 Hz. However, in both 240 Hz and 1000 Hz, the stress appears more localized in the site of the application.

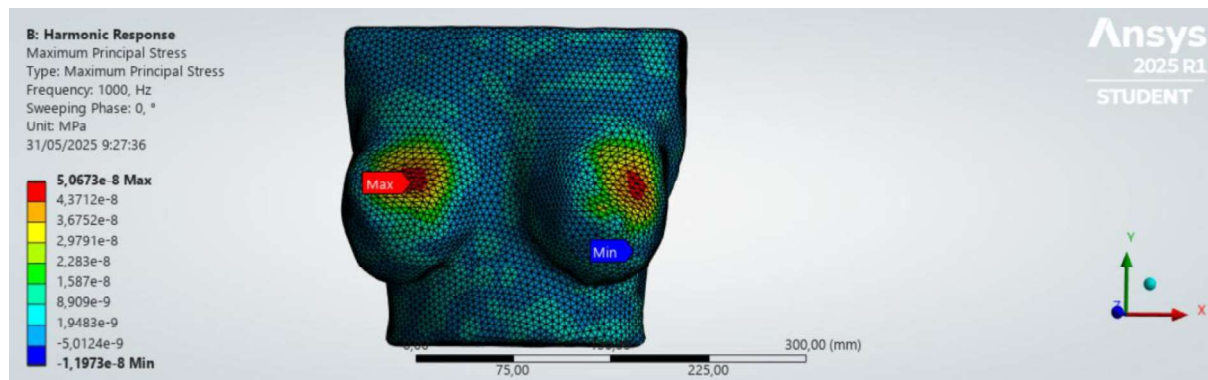


Figure 27. Maximum principal stress of the two-breast model at 1000 Hz.

5.1.5 Fluid Flow Of Milk (Simplified Geometry)

In this section, the results from the CFD simulation of the passive milk flow through a simplified circular pipe are presented.

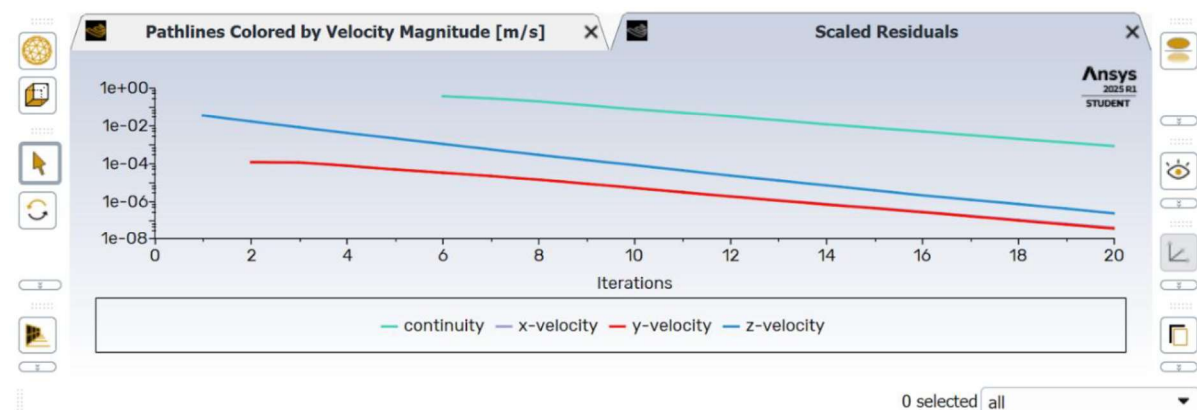


Figure 28. Convergence history of scaled residuals during CFD simulation.

The scaled residuals for continuity and the velocity components (x, y, and z) over 20 iterations of a CFD simulation simulating passive milk flow through a simplified cylindrical mammary duct are shown in Figure 28. The error between successive solutions of the governing equations is represented by residuals in computational fluid dynamics; as they gradually decrease, the solution is stabilizing and getting closer to convergence.

All of the residuals in this instance exhibit a steady downward trend, indicating that the accuracy of the numerical solution is rising.

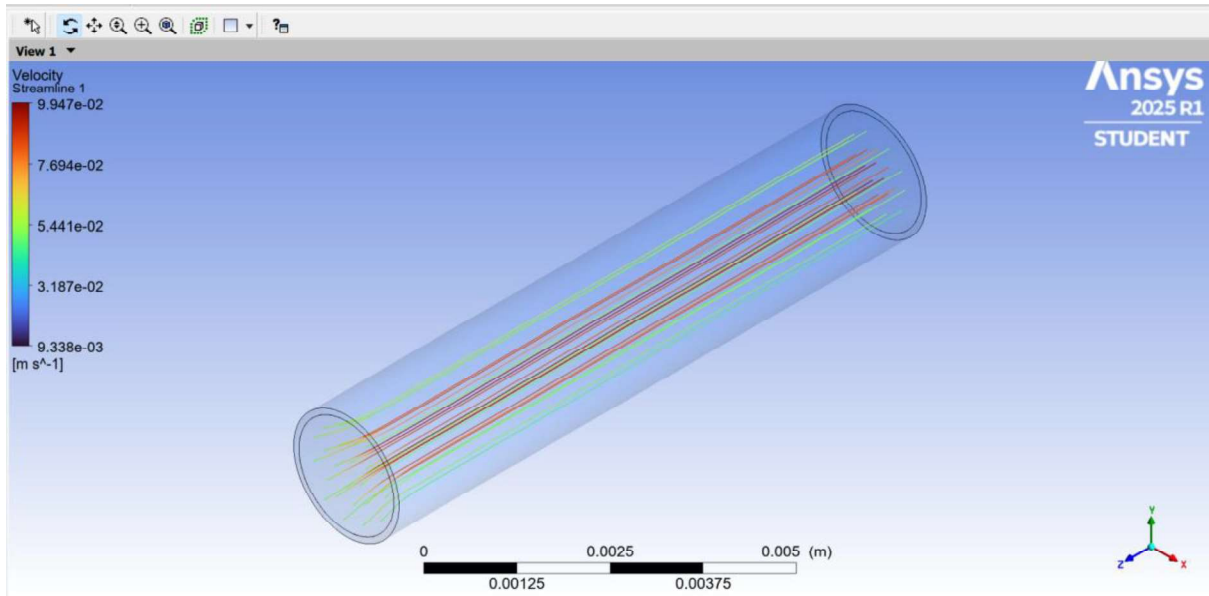


Figure 29. Streamlines flowing across the duct.

This figure shows smooth, laminar flow with streamlines evenly spaced throughout the duct. In line with a parabolic velocity profile anticipated in internal pipe flow because of the no-slip condition at the walls, the highest velocities (shown in red) are found close to the duct's center, while the flow slows down toward the wall boundaries (shown in blue/green).

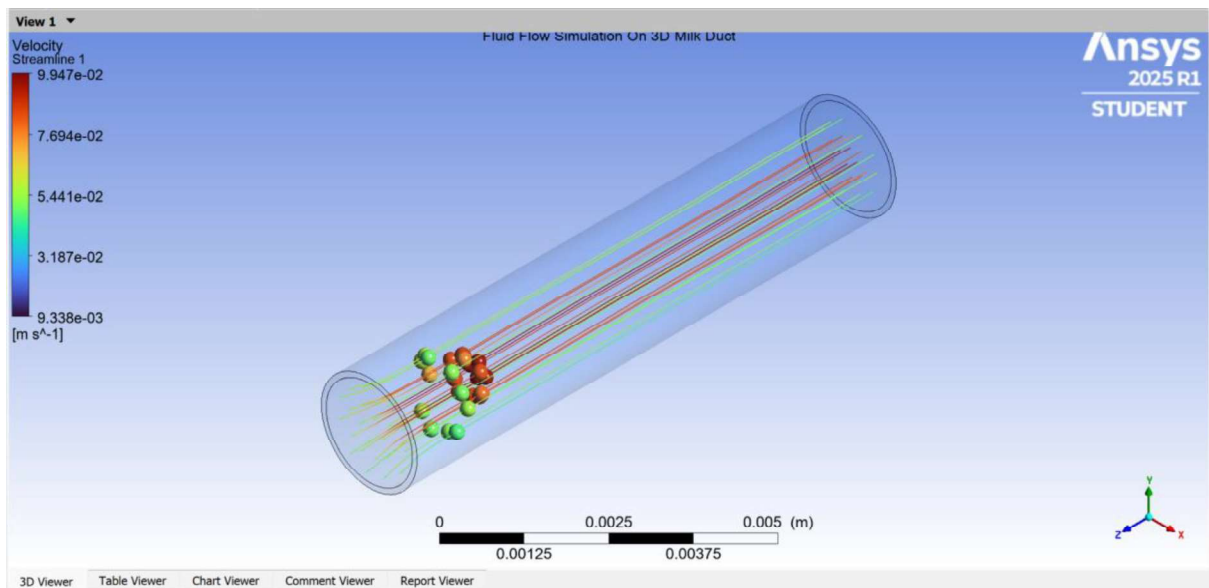


Figure 30. Velocity pathlines with marker spheres.

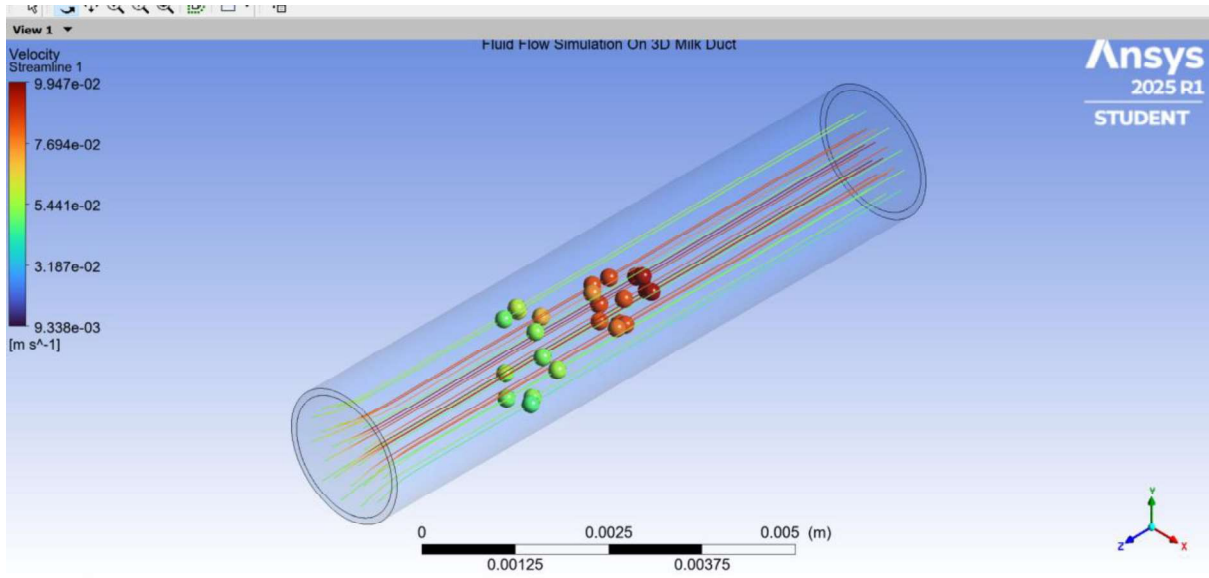


Figure 31. Velocity pathlines with marker spheres.

The velocity magnitude plots in Figure 30 and Figure 31 confirm that the inlet pressure and viscosity values chosen for milk produced realistic flow characteristics. Peak velocities remained below 0.1 m/s.

5.1.6 Fluid Flow Of Milk (Ductal Geometry)

The milk flow through a ductal structure, which resembles a mammary duct bifurcation, is shown in Figure 32 in a streamline visualization. Velocity magnitudes are indicated by the color gradient; the highest values are around 0.080 m/s.

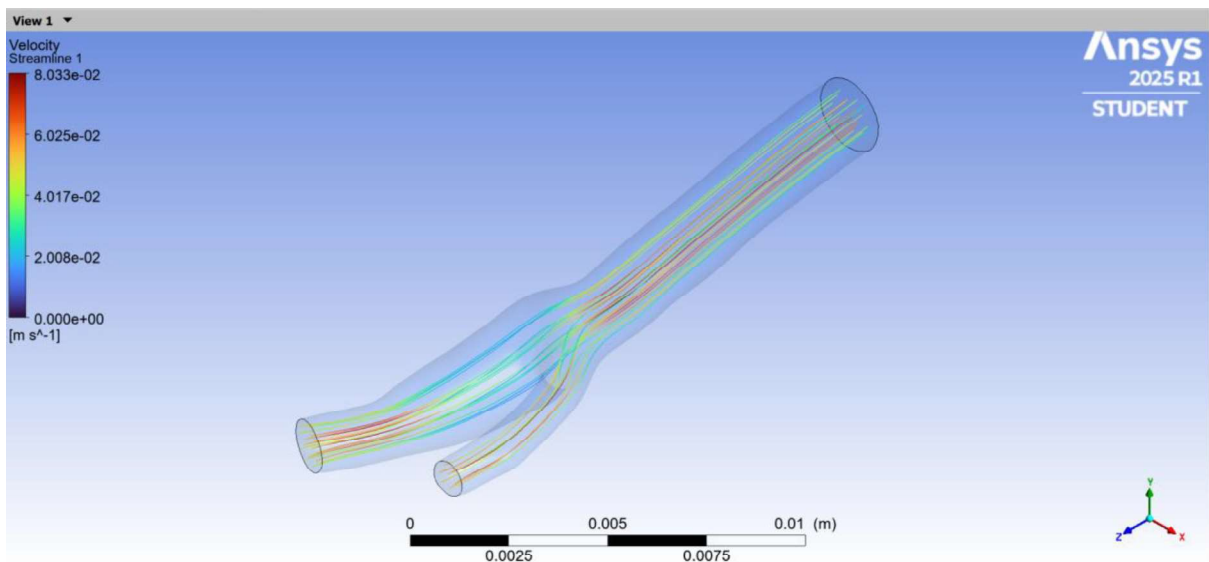


Figure 32. Milk flow through a ductal structure.

By adding particle tracking, visualized as colored spheres along the flow lines, Figure 33 is produced.

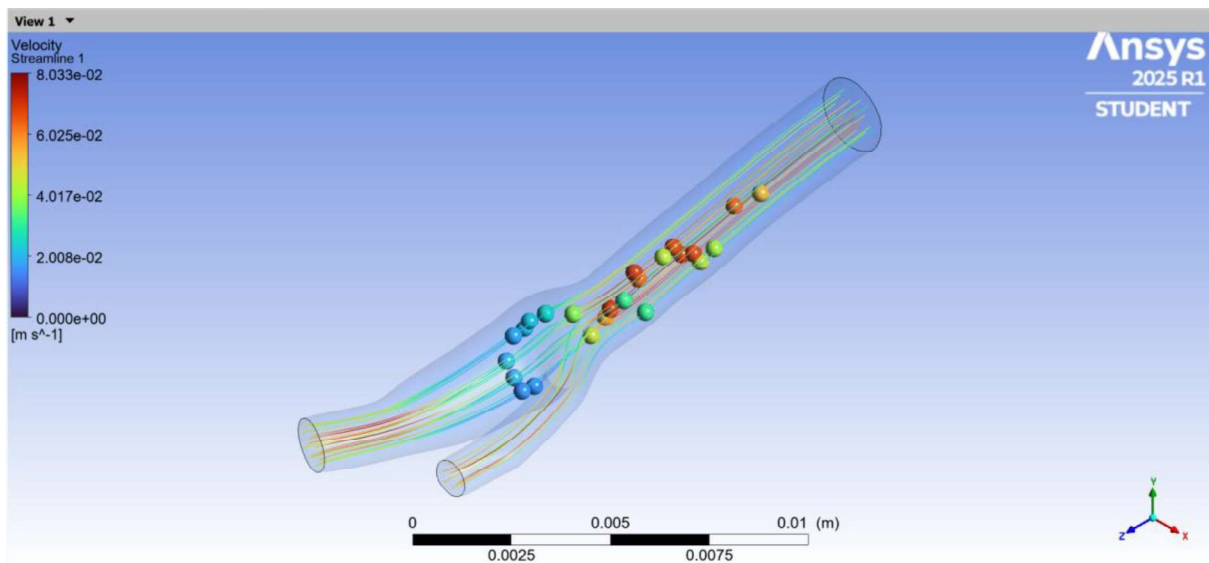


Figure 33. Milk flow through a ductal structure (with particle tracking).

5.2 Bone

5.2.1 Human Femur

5.2.1.1 Simple Geometry

5.2.1.1.1 Modal Analysis

The extraction of the first 25 modes revealed a maximum frequency value of 10,114 Hz in the 25th mode. This contrasts with previous simulations where modes were very low.

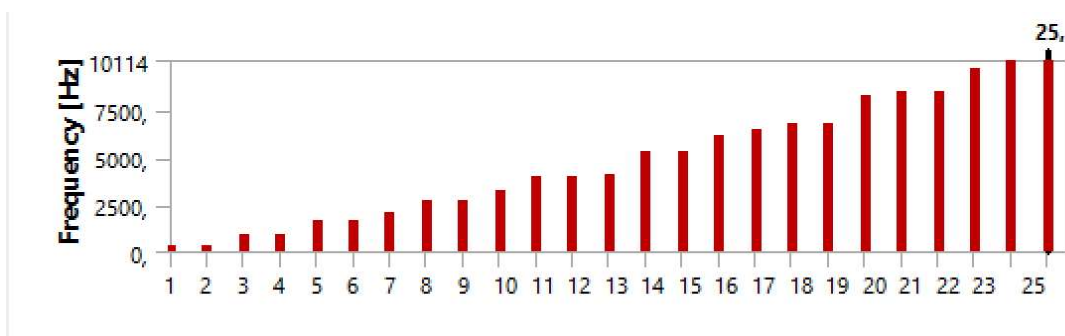


Figure 34. Frequency values achieved in each extracted mode.

Mode	Frequency (Hz)
1	329.78
2	329.88

3	891.21
4	891.48
5	1703.6
6	1704.0

Table 3. First 6 modes.

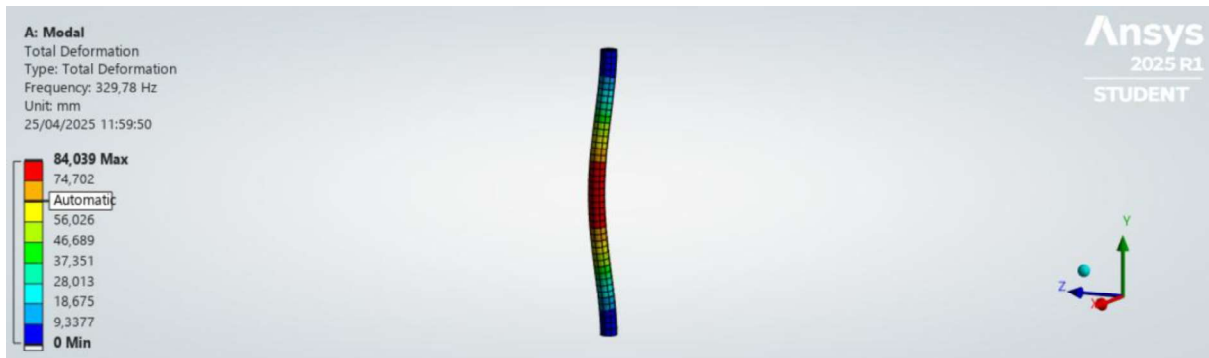


Figure 35. Total deformation of the first mode of the femur-like geometry at 329.78 Hz.

From Figure 35, it can be ascertained that the total deformation peaks at 84.039 mm, and the first mode displayed corresponds to a natural frequency of roughly 329.78 Hz.

5.2.1.1.1 Harmonic Analysis

The harmonic response of a femur-like geometry vibrating at 10,200 Hz is shown in Figure 36. The periodic wave-like pattern of the stress distribution along the length of the structure supports a higher-order mode. The peak stress is approximately 0.026 MPa, with tensile and compressive stress regions alternating. The shape is suggestive of a mode existing exactly at 10,200 Hz, which vibrates following that twisty shape.

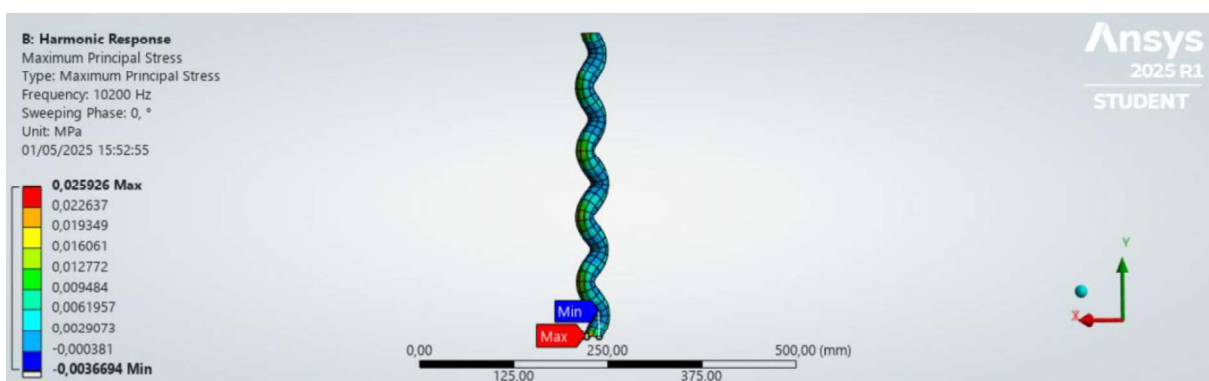


Figure 36. Maximum principal stress for a femur-like geometry at 10,200 Hz.

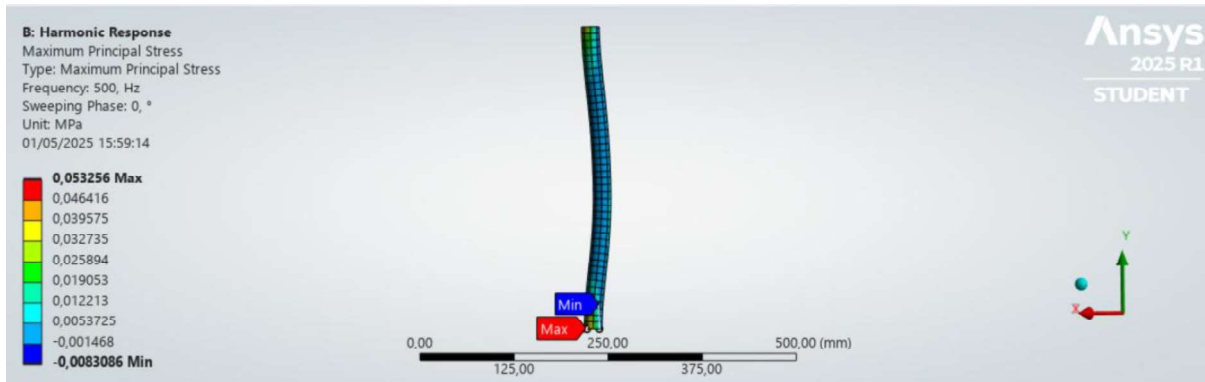


Figure 37. Maximum principal stress for a femur-like geometry at 500 Hz.

At lower, more moderate frequency values (500 Hz), where the applied load causes localized bending, the stress distribution is concentrated near the base of the model. The maximum principal stress reaches approximately 0.053 MPa, which may indicate a stronger mechanical response than at higher frequencies (10,200 Hz), possibly due to its proximity to another resonant mode.

5.2.1.2 Realistic Geometry

5.2.1.2.1 Modal Analysis

The global deformation throughout the entire femur is depicted in Figure 38. Although a general curvature along the midshaft can be seen, the deformation pattern is not immediately apparent at this scale because of the fineness of the mesh and the comparatively small displacements involved in modal analysis. A closer look at the same mode shape is provided in Figure 39, which emphasizes the area of maximum deformation surrounding the midshaft.

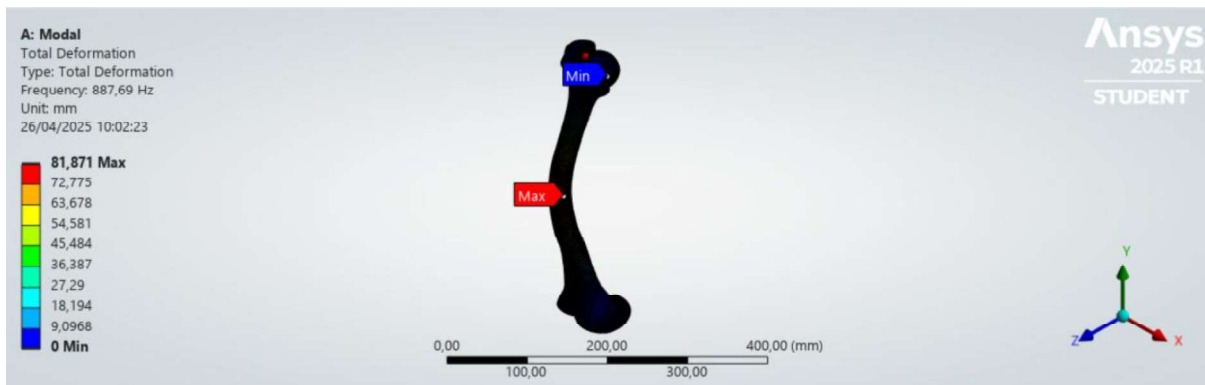


Figure 38. Total deformation of the first extracted mode. Due to the fineness of the mesh, visibility is limited.

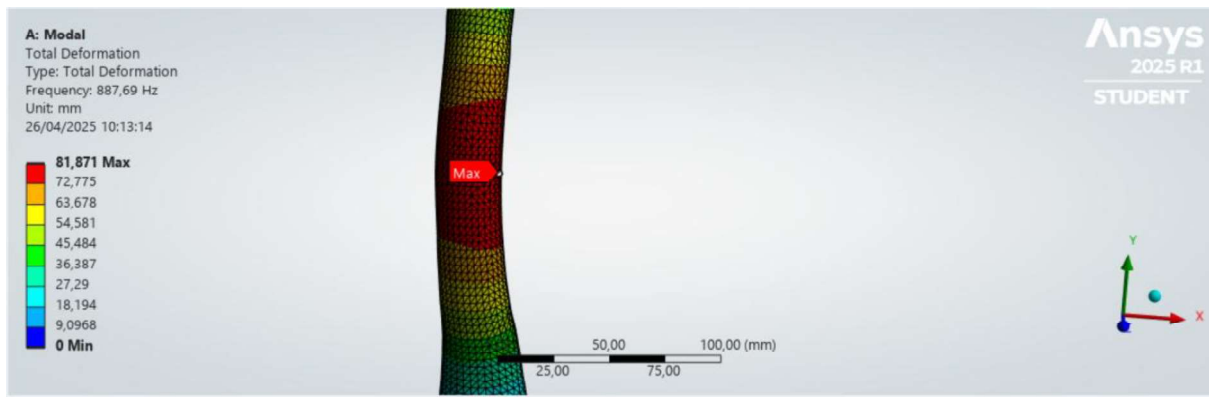


Figure 39. Total deformation of the first extracted mode. Close-up.

5.2.1.2.2 Harmonic Analysis

Near the proximal head of the femur, the maximum principal stress peaked at approximately 0.13678 MPa. The stress distribution gradually contracted toward the midshaft and distal regions, with the minimum principal stress measured at approximately -0.03216 MPa.

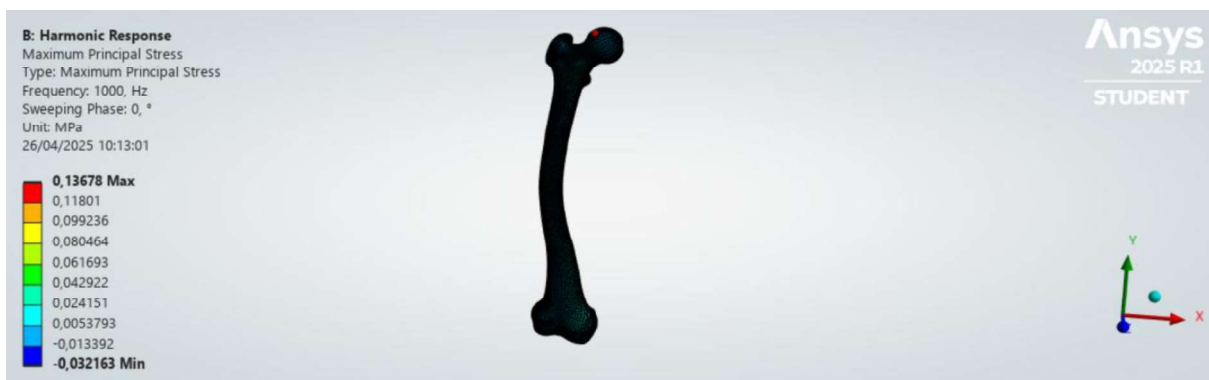


Figure 40. Maximum principal stress of the femur.

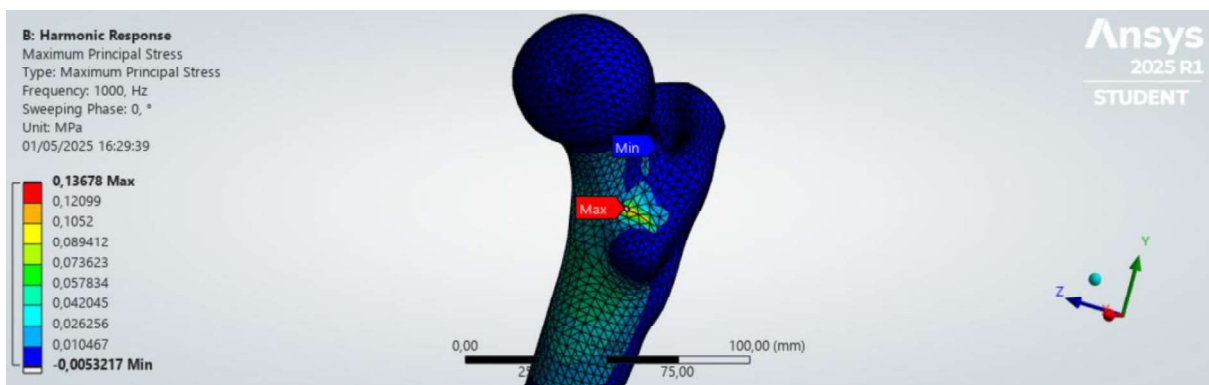


Figure 41. Maximum principal stress of the femur. Close-up.

5.2.2 Knee

5.2.2.1 Modal Analysis

The first mode, which was displayed at 89.984 Hz, had a maximum total deformation of about 38.03 mm, with the majority of the deformation occurring in the joint area. Symmetrical lateral movement in the deformation pattern was indicative of bending or rotational behavior close to the femoral-tibial interface. The last extracted mode (the sixth one) was equal to 639 Hz.

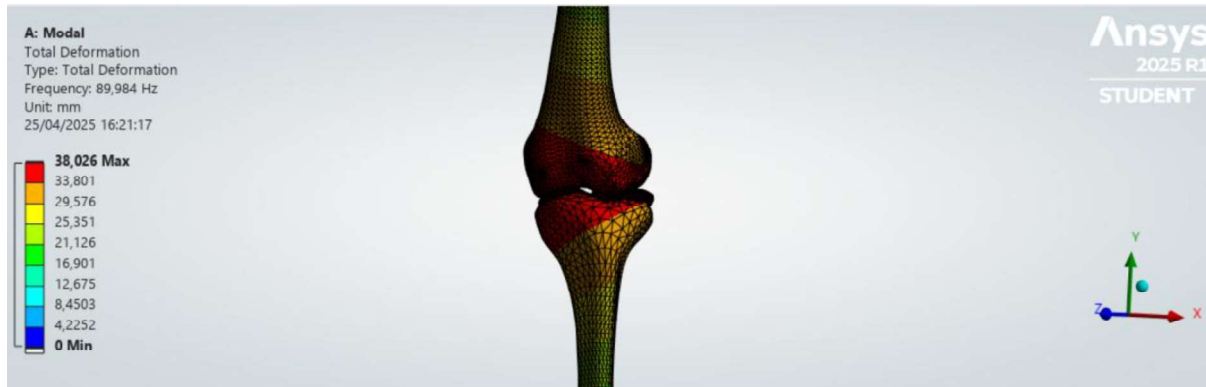


Figure 42. First mode's total deformation.

5.2.2.2 Harmonic Analysis

With a maximum principal stress of roughly 0.067 MPa, Figure 43 displays the knee's overall stress distribution. The detailed close-up view emphasizes that this peak value is located close to the knee joint, around the distal femur.

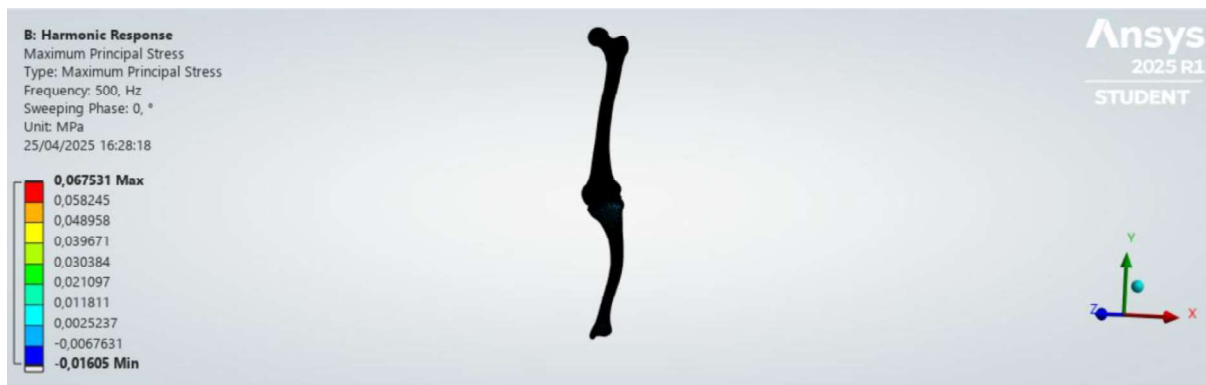


Figure 43. Maximum principal stress of the knee at 500 Hz.

The frequency of 500 Hz was chosen because it is frequently utilized in LIV and is within the range of the knee's first six natural modes.

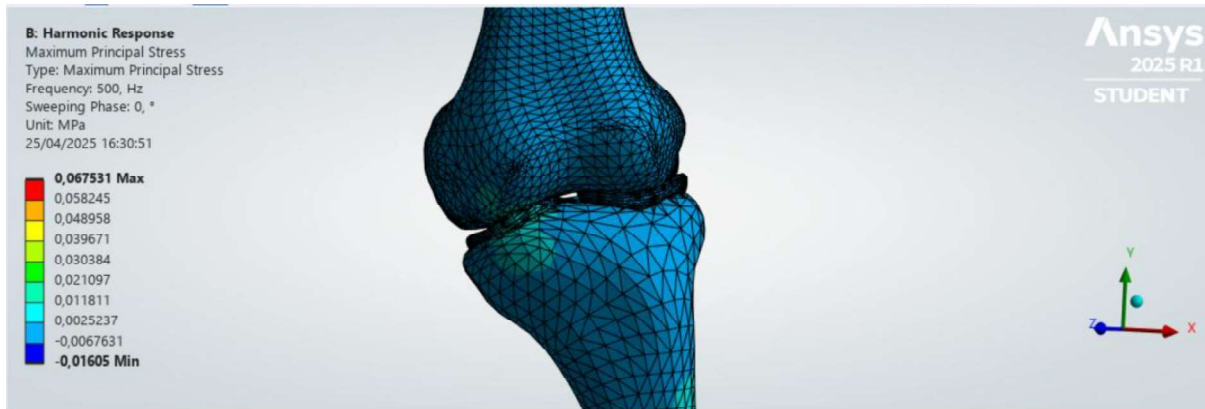


Figure 44. Maximum principal stress of the knee at 500 Hz. Close-up.

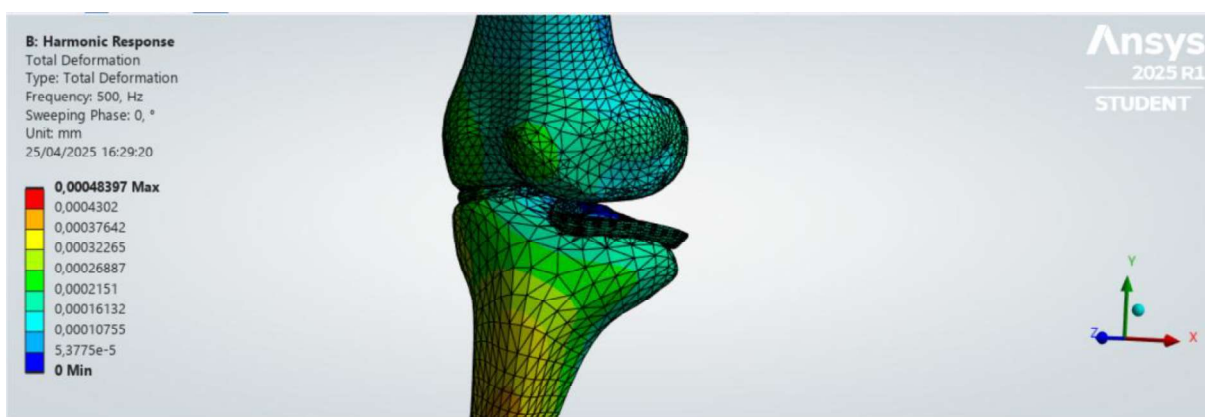


Figure 45. Total deformation of the knee at 500 Hz.

The total deformation plot shows that the displacement magnitudes are very small, with a maximum deformation of about 0.00048 mm. The deformation shows a smooth gradient, and higher values are found close to the stress concentration.

5.3 Ovary

5.3.1 Modal Analysis

When the initial modal extraction was performed using only 25 modes, the frequency range extended only up to 26.437 Hz. The number of extracted modes was then increased to 1000. Even with 1000 modes, the highest mode value was 137.25 Hz. The mode results were extracted, resulting in Figure 46 displayed below.

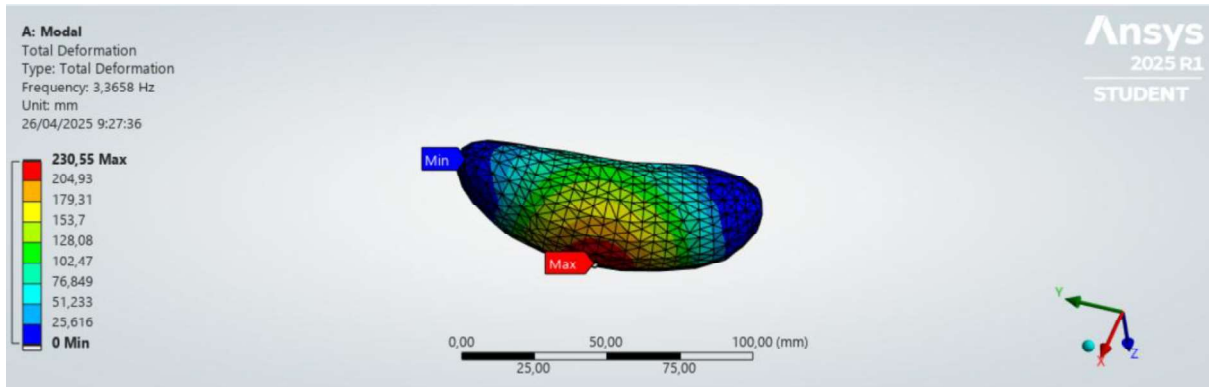


Figure 46. Total deformation of first mode.

5.3.2 Harmonic Analysis

The total deformation plot showed a maximum displacement of approximately 0.0097 mm, with the peak located near the center of the anterior surface. Deformation gradually decreases toward the posterior regions, as indicated by the color gradient.

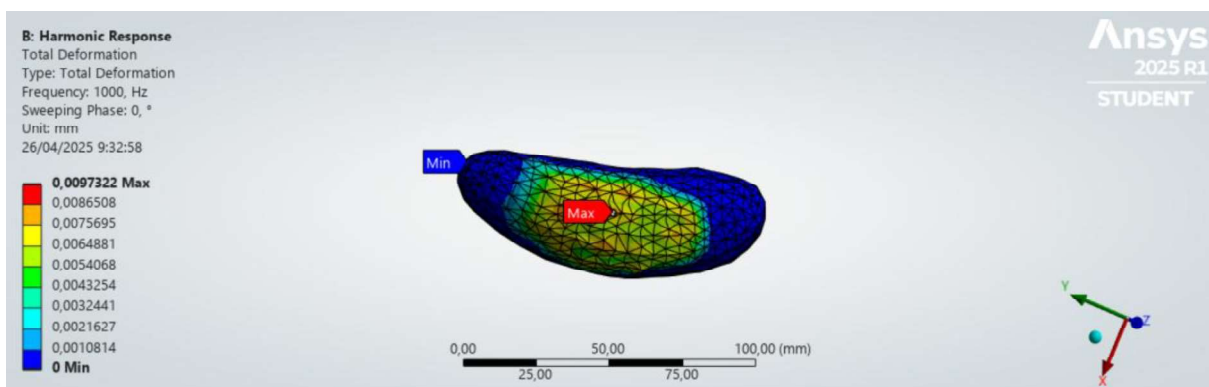


Figure 47. Total deformation of the ovary under 1000 Hz.

The simulation showed a peak value of 3.74×10^{-3} kPa for the maximum principal stress, which was also concentrated close to the tissue's center. Lower values are seen at the lateral and posterior ends of the stress field, which is distributed more widely than the displacement. The lowest stress measured was -4.84×10^{-4} kPa.

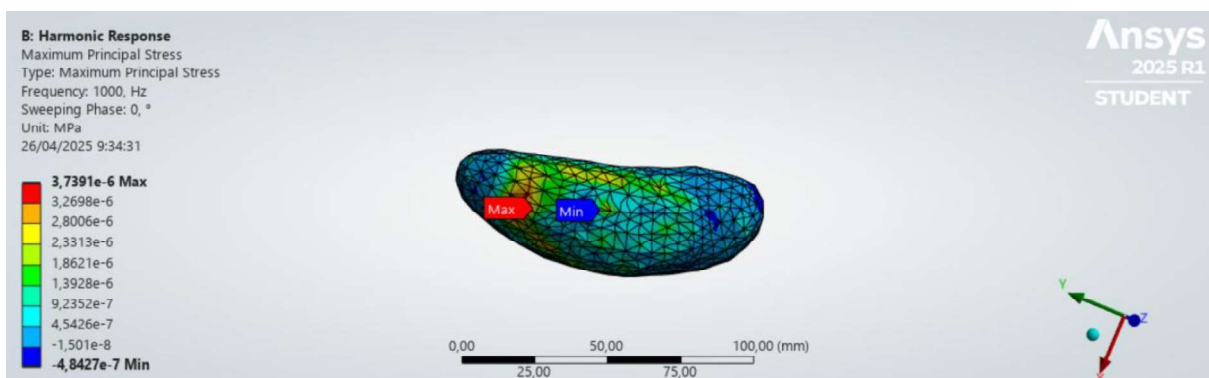


Figure 42. Maximum Principal Stress of the ovary under 1000 Hz (first subrange)

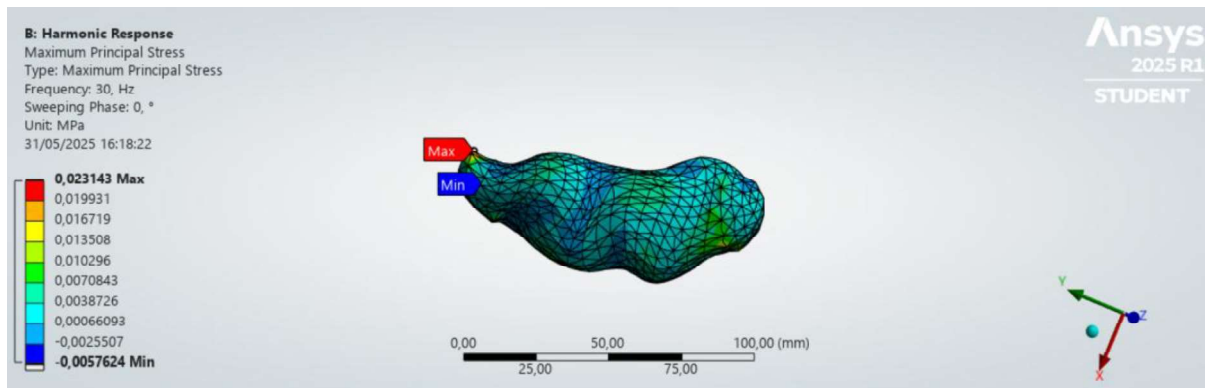


Figure 43. Maximum Principal Stress of the ovary under 30 Hz (first subrange)

In Figure 43, it's possible to observe that at 30 Hz, a maximum principal stress of approximately 0.0231 MPa is reached, concentrated at a small region near one end of the tissue.

6. Discussions

In this section, the previously obtained results will be discussed, as well as the ethical-social impact, sustainability, and diversity aspects applicable to this thesis.

6.1 Discussion of the results

6.1.1 Breast and Lactation

6.1.2.1 Breast

The fat tissue cube's modal and harmonic analyses showed distinct resonance effects, especially at 30 Hz, which is one of the cube's inherent frequencies. The structure showed enhanced mechanical responses, with large displacements and high stress levels, when the excitation frequency matched this mode. It was shown that resonant amplification significantly controls mechanical behavior under harmonic loading, producing stresses that are significantly higher than those observed under non-resonant conditions. The distinctive wavy, buckled deformation pattern matched the spatial mode shape at 30 Hz.

The observed stresses under LIV were very low when the breast-like geometry was modeled using the properties of adipose tissue. This was in line with the expected mechanical response of soft biological tissues under normal physiological loading.

A maximum principal stress of roughly 0.106 kPa was found for the breast tissue model using harmonic response analysis at 1000 Hz. The model showed a forced response even though the excitation frequency was higher than the system's last extracted modal frequency (~437 Hz). This suggests that external LIV can produce mechanical fields strong enough to support biological processes like mechanotransduction without running the risk of mechanical damage. The wide, low-magnitude distribution of stress points to the formation of a mechanical environment that may be advantageous for physiological processes like improved blood flow and lymphatic drainage.

In contrast to the 1000 Hz (~0.106 kPa) and 10,000 Hz (~0.001 kPa) cases, the 40 Hz harmonic response produced noticeably higher stresses (~2.514 kPa). This is consistent with the dynamic behavior of soft tissues, where stress distributions are amplified by lower-frequency stimulation that causes large-scale deformations and dominant modal responses. These results support broader evidence that low-frequency mechanical inputs more effectively trigger biological effects like fluid movement, cellular signaling, and mechanotransduction. They also show that lower-frequency LIV stimulation is significantly more effective at mechanically activating soft tissues.

For the two-breast model, harmonic analysis similarly showed that lower-frequency excitations (like 20 Hz) induced higher internal stresses than higher-frequency loads (240–1000 Hz). This is explained by the engagement of whole-structure deformations and dominant low-frequency modes (with the highest extracted mode at 18.578 Hz), which amplify stress across the model. This explains why, even in an undamped system, the breast model experiences greater mechanical activation under low-frequency harmonic loading.

6.1.2.2 Lactation

The results of the CFD simulation demonstrated a well-developed laminar flow regime in a simplified cylindrical model of a mammary duct, which replicates passive milk transport under continuous, pressure-driven conditions. As anticipated in low Reynolds number flows, peak velocities along the central axis progressively dropped toward the duct walls, creating a parabolic velocity profile. And with spherical markers, the particle pathlines demonstrated the uniformity of the flow by confirming that particles near the boundaries moved slower than those near the center.

To validate the models, reported lactational milk flow rates (0.12–0.72 g/s; Prime et al., 2009) were converted to velocity units using the formula $v = m / (\rho \cdot A)$, where m is the mass flow rate (converted to kg/s), ρ is the density of milk (1030 kg/m³), and A is the area of a 2 mm-diameter duct. This yielded an estimated velocity range of 0.037–0.222 m/s, placing the simulations' peak velocity (just under 0.1 m/s) comfortably within the physiologically expected range, this way supporting the validity of the setup.

A useful baseline for comprehending passive milk transport under physiological conditions was created by these specific CFD models, which effectively captured the merging flow dynamics at both simple ducts and at duct bifurcations, using realistic duct geometry. These results lay the groundwork for future fluid-structure interaction (FSI) simulations that could examine whether LIV can improve ductal drainage, lessen blockages, or alter mechanotransduction processes within the mammary tissue. Future clinical approaches to promoting lactation health may benefit from these insights.

6.1.2 Femur

The harmonic response analysis revealed a distinct wavy deformation pattern at 10,200 Hz, which was indicative of higher-order mode shapes found in high-frequency dynamic systems. Even though it corresponds with the 25th extracted mode, this shows that even rigid materials, such as cortical bone, display intricate vibrational shapes when excited

at high frequencies. Crucially, this does not mean that the femur exhibits flexible behavior. Instead, these patterns, which have several nodes and antinodes, are the result of the superposition of natural modes that were extracted during modal analysis. The actual displacements are small, even though ANSYS visually magnifies these deformations for clarity.

Different mechanical behaviors under LIV were revealed by simulations run at 500 Hz and 10,200 Hz. At 500 Hz, the highest stress response was observed, most likely as a result of strong resonant amplification close to the bone's natural frequency range. The midshaft showed a smooth distribution of stress, indicating mechanical stimulation and efficient vibrational coupling. Despite being close to a resonant peak, stress decreased to about 0.026 MPa at 10,200 Hz. In spite of being close to a natural mode, the mechanical filtering effect of bone at high frequencies, where the material's inertia limits the amplitude of vibration, can reduce stress buildup.

Interestingly, the realistic femur model consistently showed higher stress concentrations than the simplified cylindrical model, highlighting the importance of anatomical features in amplifying mechanical loads. Vibrations at 1000 Hz produced resonant effects (first mode was around 800 Hz) with elevated stress responses (~ 0.13678 MPa), suggesting that matching LIV device settings proximal to that range could enhance mechanical stimulation for applications like osteoporosis prevention or fracture healing. It could also indicate an amplified risk of fracture, so testing the LIV frequencies in a clinical setting is of paramount importance. The anatomical complexity of the realistic femur, which includes curvature, varying cross-sections, and prominent landmarks like the femoral head and condyles, introduced stress concentrators and supported more complex vibrational modes than the idealized cylindrical model. This increased local stress under harmonic loading.

6.1.3 Knee

Modal analysis identified key natural frequencies in the low- to mid-frequency range, notably around 90 to 600 Hz, where deformation was concentrated at the femoral-tibial interface. This suggests that vibrational stimulation within these frequency bands could effectively activate the knee's natural resonant modes.

The deformation patterns observed, marked by symmetrical lateral displacement and bending at the joint, suggest efficient energy transfer to the articular surfaces and surrounding bone structures. This kind of mechanical engagement could have important clinical benefits, potentially helping to improve joint function, stimulate local mechanotransduction pathways, or support recovery after injury or degenerative changes. By pinpointing the most mechanically sensitive areas, it becomes possible to design targeted LIV protocols that deliver localized therapeutic effects without placing excessive strain on the tissue, ultimately promoting better joint health.

While overall stress and deformation were low, the 500 Hz harmonic response showed that local stress concentrations (~ 0.067 MPa) appeared close to the distal femur, particularly around the knee joint. This is explained by the intricate geometry and interfaces of the area, which by nature serve as stress concentrators even when subjected to mild vibrational inputs.

Even though the maximum deformation was minimal (~ 0.00048 mm), these little mechanical clues might still have biological significance. The hypothesis that LIV may be a promising adjunct in musculoskeletal rehabilitation is supported by these findings. It targets joint regions that are susceptible to degeneration or that need increased mechanical stimulation in order to facilitate healing and functional recovery.

6.1.2 Ovary

An understanding of how soft reproductive tissues react to LIV was gained from the ovary simulation. The modal analysis showed that the ovary has very low natural frequencies compared to stiffer tissues like bone. Even after 1000 modes were extracted, the first modes only reached about 137 Hz. This illustrates the soft-tissue, compliant nature of the ovary, which is marked by high deformability and low stiffness. Its biological function as a soft, vascularized endocrine organ, where mechanical stimuli can spread widely but are unlikely to produce sharp local stress concentrations, is consistent with this mechanical behavior.

Maximum principal stresses of roughly 0.0231 MPa were revealed by harmonic response analyses, especially at 30 Hz (amplified due to resonance). The simulations showed lower levels of stress and deformation at higher frequencies (up to 1000 Hz), indicating that ovarian tissue is sensitive to resonance and that LIV devices for soft tissue should be designed taking this into account. This mechanical response suggests that externally applied vibrations can introduce mild mechanical signals that may affect cellular mechanotransduction, lymphatic drainage, or local microcirculation even in the absence of resonant amplification.

Crucially, these results demonstrate that ovarian tissue can effectively tolerate the mechanical conditions created by LIV, paving the way for further investigation into the potential benefits of such stimulation for gynecological health, such as improving perfusion or treating diseases like cystic blockages or PCOS.

6.2 Ethical-Social Impact, Sustainability, and Diversity Aspects

6.2.1 Ethical Considerations

This thesis does not directly experiment on humans or animals; instead, it is based on computational modeling. Therefore, neither informed consent nor ethical approval were needed. But the biological phenomena that are modeled, like the response of breast tissue to vibration, lactation physiology, and musculoskeletal biomechanics, are sensitive by nature, especially when it comes to the health of women. The use of simulation to circumvent potentially intrusive or uncomfortable experiments has important ethical ramifications. An ethical substitute for animal testing is provided by computational methods, which also spare human subjects from pain or danger, particularly in situations like breast vibration studies where experimental methods may be intrusive or scary.

6.2.2 Social Impact

This thesis's research has the potential to significantly advance rehabilitation and healthcare. This work contributes to the development of safer, non-invasive treatments for diseases like osteoporosis, milk stasis, and joint degeneration by creating computational

models that mimic how tissues react to LIV. These illnesses frequently need long-term care and impact a wide range of people, including the elderly, new mothers, and those with limited mobility. For vulnerable populations that are less suited for high-impact physical therapy, LIV-based therapies may provide easily accessible, affordable substitutes or supplements to pharmaceutical and surgical treatments. This establishes the research as a significant contribution to raising well-being, encouraging independence, and improving quality of life.

However, if new technologies are not distributed fairly or are not created with inclusivity in mind, they may inadvertently make health disparities worse. For example, LIV devices may only benefit certain social groups if they become commercially successful but are still too expensive or unavailable for lower-income communities. To guarantee widespread societal benefit, consideration must be given to the development, pricing, and application of such LIV devices.

6.2.3 Sustainability

Computational simulation offers a low-resource substitute for physical prototyping and repeated experimentation from a sustainability perspective. Researchers can drastically cut down on material waste, energy use, and environmental impact by employing numerical techniques like CFD and FEA instead of traditional trial-and-error testing. Furthermore, compared to more energy-intensive rehabilitation technologies, LIV devices themselves are probably going to have a lower environmental impact because they are mechanically straightforward and non-invasive. Future research should take into account end-of-life recyclability, energy efficiency during operation, and the sourcing of materials for device production.

6.2.4 Diversity and Inclusivity

By choosing applications that are pertinent to a broad spectrum of people, across genders, age groups, and medical conditions, this study was created with inclusivity in mind. One area of biomedical engineering that has historically been underrepresented is the modeling of lactating breast tissue, which fills a major health gap for women. While joint and spine modeling broadens the applicability of this work to individuals with chronic musculoskeletal conditions or mobility impairments, the inclusion of simulations related to osteoporosis acknowledges the aging population. However, there is potential for further growth. The inclusivity and applicability of these models would be further improved by including population-specific anatomical and physiological data (such as across ethnic groups, age ranges, or body types). Simulation frameworks like the ones created in this thesis could be modified to support customized treatment planning as personalized medicine gains traction.

6.3 Limitations

It's critical to recognize the limitations of this thesis. First, because of its numerical problem size cap, the ANSYS Student version placed stringent restrictions on model complexity. This limitation capped the resolution and detail of vibrational simulations by making full-method harmonic analyses with a finely meshed model impracticable.

Furthermore, more complex or anatomically accurate material models (especially those involving viscoelastic or layered structures) could not be incorporated due to the software's limitations on material definition. Material models were created from scratch using defined values for density and isotropic elasticity, since the free ANSYS Student license does not allow the use of predefined hyperelastic models, like the Neo-Hookean formulation.

More complex simulations, like multi-physics coupling or fully time-dependent analyses, were not possible due to license limitations. The inability to use CFD to model how milk flow responds to LIV is one important consequence of this. The scope of this project did not allow for the development of a user-defined function (UDF), which would be required to impose harmonic loading in a CFD environment.

Furthermore, due to a lack of directional data and the additional complexity required for such modeling, orthotropic elasticity, which more accurately depicts the directional dependence of mechanical properties, was not used in the modeling of bone tissue. Bone was thus reduced to an isotropic substance. Despite its importance in accurately representing the dynamic response of biological tissues, damping was also left out of the models in favor of computational simplicity.

Finally, because no experimental validation was done, the study is totally dependent on parameter values found in the body of existing literature. Although a great deal of work was done to choose data that was widely accepted and physiologically plausible, this dependence reduces the accuracy and clinical relevance of the results. The results of this thesis should therefore be viewed as exploratory in nature, providing theoretical insights rather than conclusive findings.

7. Conclusions

This thesis used FEA and CFD to investigate the mechanical responses of various biological tissues to LIV. This work demonstrated the distinct mechanical signatures that each tissue demonstrates under vibrational loading using a wide range of models, including adipose tissue, breast tissue, mammary ducts, femur, knee joint, and ovary. These models provided important understanding for possible biomedical devices.

One of the primary findings is that vibrational response is largely determined by geometry. Results differed greatly based on the shape and anatomical complexity of the model, even with the same material parameters. In contrast to anatomically realistic models such as the femur and knee, which displayed local stress amplification at curved or transitional regions due to anatomical landmarks and geometry-induced stress concentrations, simplified geometries such as cubes or cylinders displayed different resonant peaks and distributed stress patterns. This demonstrates how crucial it is to take into account both geometrical and material considerations when developing and implementing LIV treatments.

Soft tissues with low stiffness are especially sensitive to low-frequency stimulation (20–40 Hz), as shown by the breast models, which exhibit increased stress and deformation close to their natural modes. Nonetheless, tissues showed smooth, low-magnitude stress gradients even at higher frequencies (up to 1000–10,000 Hz), suggesting mechanical stability under LIV

device conditions. These findings imply that LIV may be able to safely promote soft tissue function, potentially improving cellular mechanotransduction, lymphatic drainage, or blood flow without resulting in mechanical damage.

Simulations of bones and joints also demonstrated that while higher-frequency excitation was filtered by the material's inertial characteristics, mid frequencies (especially those that matched or slightly exceeded natural frequencies) produced significant mechanical responses, which may be therapeutic or even dangerous. The midshaft of the femur has repeatedly been identified as a mechanically sensitive location, with both realistic and simplified models demonstrating increased stresses under resonant circumstances. The notion that LIV may have potential uses in bone remodeling, joint health, or rehabilitation techniques is supported by the knee joint's similar display of local stress concentrations near the distal femur at frequencies of about 500 Hz.

Furthermore, the results support the idea that gentle vibratory stimulation might present opportunities for gynecological interventions, pending further research. The ovary, a soft, vascularized tissue, displayed very low natural frequencies, and stress responses remained modest under LIV conditions.

All things considered, this thesis shows that LIV-induced mechanical stimulation can be replicated in a variety of biological tissues, offering a useful foundation for creating non-invasive treatment plans. These simulations provide a theoretical basis for fine-tuning LIV device parameters to target particular tissues or clinical objectives by mapping resonant modes, stress distributions, and deformation patterns.

Despite software and modeling limitations, the study opens promising avenues for future work. In the end, the results establish the foundation for upcoming advancements in biomedical engineering and therapeutic device development by reiterating the vital significance of meticulous, tissue-specific simulation when investigating the mechanobiological effects of LIV.

8. Future Work

This thesis covered a wide range of biological tissues through simulations. It is crucial to recognize, though, that a lot of these simulations had to be made simpler because of software or computational constraints. For example, the knee model solely focused on the bony structures, purposefully excluding soft tissues and cartilage. Creating comprehensive models that include all anatomical components would provide a more thorough understanding of joint mechanics under LIV in future work where computational efficiency is less of a constraint.

To better understand how harmonic loading impacts milk flow dynamics, future studies could build on the lactation simulations by incorporating simplified vascular models or using fluid-structure interaction (FSI) techniques. Furthermore, by incorporating layered tissue structures (such as skin, adipose tissue, and glandular tissue) instead of presuming tissue homogeneity, future models could enhance biological realism and offer a more nuanced perspective of mechanical response and stress distribution.

Future studies should ideally include in vivo measurements or physical testing to impulse the validity of the conclusions made from these computational models. Such experimental validation would guarantee that the modeled responses accurately reflect actual tissue behavior under LIV and help validate the accuracy of the simulation results.

Incorporating damping effects and soft tissue heterogeneity into the simulations would also help to provide a more complete understanding of tissue response. The baseline data generated here provides the foundation for building more intricate models for the ovarian models. To more accurately evaluate LIV's therapeutic potential in reproductive medicine, these might include FSI or effects on hormone regulation.

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