

**HUMAN NECK DYNAMIC MODELING AND DESIGNING
ASSISTIVE NECK DEVICE DESIGN FOR NECK PAIN
PREVENTION**

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SEPTEMBER, 2025

HUMAN NECK DYNAMIC MODELING AND DESIGNING ASSISTIVE NECK DEVICE DESIGN FOR NECK PAIN PREVENTION

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Thesis
Submitted in Partial Fulfillment of the
Requirements for the Degree of

Master of Science

in
Mechanical Engineering

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September, 2025

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To my mom, for all the good memories...



DECLARATION OF ORIGINALITY

I hereby declare that I am the sole author of this thesis and that this is the true copy of my thesis, including the final revisions, approved by my thesis committee. All data and information have been obtained, produced, and presented in accordance with the rules of research ethics and principles of academic honesty. As required by these rules, to the best of my knowledge I have acknowledged ideas, thoughts, and any copyrighted material in accordance with the standard referencing rules. I certify that any part of this thesis has not been submitted for a degree or diploma in another educational institution.

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ABSTRACT

Pain in the neck is a frequent musculoskeletal issue, particularly among office employees, those who remain standing for extended durations, and individuals who hold static positions for a long time. Continuous forward neck flexion leads to muscle fatigue, elevated EMG activity, and decreased pain tolerance, negatively impacting quality of life and productivity. This research presents NACK (Neck Assistive Cervical Kit) — an aid created with a comprehensive multibody neck model to assist in preventing neck discomfort by promoting proper posture. The system integrates a helpful headband with a conventional office chair to reduce excessive neck bending. A 6-DOF biomechanical neck model was created in MSC ADAMS to replicate realistic neck motions. The optimal placement and stiffness of the spring were identified using Design of Experiments (DOE). The ultimate design was enhanced in SOLIDWORKS, evaluated with finite element analysis (FEA), and prototyped using 3D printing for weight and comfort. Twenty participants underwent tests using surface EMG sensors placed on the sternocleidomastoid (SCM) and upper trapezius (UT), along with IMU data. Findings indicated diminished SCM and UT activity along with a 25% reduction in total neck muscle force, demonstrating the preventive advantage of the design. The model is capable of analyzing deep neck muscles, showcasing NACK's ability as a practical, ergonomic solution to alleviate neck strain for individuals working at desks.

Keywords: Cervical Spine, Office Ergonomics, Assistive Device, Simulation, Design

ÖZET

Boyun ağrısı, özellikle ofis çalışanları, uzun süre ayakta çalışanlar ve iş sırasında uzun süre sabit pozisyonda kalan kişiler arasında yaygın görülen bir kas-iskelet sistemi sorunudur. Uzun süreli öne eğik boyun duruşu kas yorgunluğuna, artan EMG aktivitesine ve ağrı eşiğinde azalmaya yol açarak yaşam kalitesini ve verimliliği düşürmektedir. Bu çalışma, sağlıklı duruşu destekleyerek boyun ağrısını önlemeye yardımcı olmak amacıyla geliştirilen NACK (Neck Assistive Cervical Kit) adlı yenilikçi bir boyun destek cihazını tanıtmaktadır. Sistem, aşırı boyun fleksiyonunu sınırlamak için standart bir ofis sandalyesine entegre edilmiş destekleyici bir kafa bandından oluşur. Gerçekçi boyun hareketlerini simüle edebilmek için MSC ADAMS ortamında 6 serbestlik dereceli (6-DOF) ayrıntılı bir biy-omekanik boyun modeli geliştirilmiştir. Deney Tasarımı (DoE) yöntemiyle optimum yay sertliği ve konumları belirlenmiştir. Nihai tasarım SOLIDWORKS'te detaylandırılmış, sonlu elemanlar analizi (FEA) ile test edilmiş ve hafiflik ve konfor için 3B baskı ile pro-totip üretilmiştir. 20 katılımcı ile yapılan deneylerde sternokleidomastoid (SCM) ve üst trapez (UT) kaslarına yüzey EMG sensörleri yerleştirilmiş, IMU verileri kullanılmıştır. Sonuçlar, SCM ve UT aktivitesinde azalma ve toplam boyun kas kuvvetinde yüzde 25 düşüş göstermiştir. Model aynı zamanda derin boyun kaslarının analizine de imkân tanımaktadır. Bu bulgular, NACK'in ofis çalışanları için pratik ve ergonomik bir önleyici destek çözümü olabileceğini ortaya koymaktadır.

Anahtar Kelimeler: Servikal Boyun, Ofis Ergonomisi, Yardımcı Cihaz, Simülasyon, Dizayn

ACKNOWLEDGEMENTS

I want to convey my sincere gratitude to all those who assisted and advised me during my educational journey. This thesis would not have been possible without their essential support and motivation.

I am deeply thankful to Assoc. Prof. Şendur, my supervisor, whose exceptional support, expert guidance, and extraordinary patience were vital at each phase of this research. His perceptive advice and considerate input not only influenced the trajectory of this thesis but also provided ongoing inspiration. I feel incredibly lucky to have been guided by such a committed and insightful mentor and greatly appreciate the vital role he had in the achievement of this project.

I also wish to extend my sincere gratitude to my co-advisor, Assis. Prof. Ünal, for his valuable input, constructive criticism, and for generously opening his laboratory for my experiments. His willingness to share his expertise, provide access to essential resources, and engage in productive brainstorming sessions greatly enriched this study and improved its overall quality.

I also want to express my gratitude to Nima Heidari, Ömer Sezgin, and Omid Orfae for their significant input in advancing this work. I am truly grateful to Çağan Kutlay and Melis Şelte for their generous support and help throughout the experimental phase of this study.

A heartfelt thank you to my family for their endless love, encouragement, and unwavering belief in my abilities. Their constant support has been my foundation throughout this challenging journey. I love you so much — Mom, Dad, Serra, and Elif.

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LIST OF ACRONYMS AND ABBREVIATIONS

ABS	Acrylonitrile Butadiene Styrene
AC	Attachment Concept
ALS	Amyotrophic Lateral Sclerosis
AROM	Active Range of Motion
CAD	Computer-aided Design
CES	cervical erector spinae
COM	Center of Mass
CVA	craniovertebral angle
DHS	Drop Head Syndrome
DOE	Design of Experiments
DOF	Degrees of Freedom
ECG	electrocardiogram
EMG	electromyography
FEA	Finite Element Analysis
FRP	flexion relaxation phenomenon
FRR	flexion relaxation ratio
Hz	Hertz
IMU	Inertial measurement unit
MBD	Multibody Dynamics
MSD	musculoskeletal disorders

MVC	Maximum Voluntary Contraction
NACK	Neck Assistive Cervical Kit
NBQ	Neck Bournemouth Questionnaire
PLA	polylactic acid
RMS	Root Mean Square
RoM	Range of Motion
RRS	Revolute-Revolute-Spherical joints
RULA	Rapid Upper Limb Assessment
SENIAM	Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles
SCM	sternocleidomastoid muscle
sEMG	surface electromyography
UT	upper trapezius
VNPS	Verbal Numerical Pain Scale

1. INTRODUCTION

The literature review for this study is thoroughly addressed in Section 1.1, covering different facets associated with neck injuries in office employees, designs of support systems for the cervical spine, mathematical models concerning the Cervical Spine and ADAMS, and experiments conducted on office workers. Additionally, Section 1.2 outlines the research gap and the primary objectives of this thesis. The structure of the thesis is outlined in Section 1.3, and Section 1.4 offers a guide for readers to assist them in moving throughout the document.

1.1 Literature Review

To start this part, it is better to explain what is neck pain and what is its outcome to the our daily life. To specify the neck pain region which is between the occiput and the scapulae spine was referred to as neck pain.[1]

In the situations such as long-term care for neck pain is necessary for neck pain and injuries, which places a financial and economic strain on society.[2]There are four different types of factors can be root to for the neck pain. These are:

- a) Musculoskeletal disorders: Postural abnormalities and imbalances in the cervical muscles are two main causes of neck pain. People who spend a lot of time at desktops are particularly prone to cervical muscle fatigue.
- b) Cervical disc degeneration: As people age, their cervical discs may degenerate, compressing nerve roots and causing persistent neck pain.
- c) Dysfunction of the facet joints: Degenerative alterations in the facet joints might cause neck pain since they are essential to the cervical spine's movement.
- d) Trauma and injuries: One of the most frequent causes of both acute and chronic neck pain is cervical trauma brought on by falls, sports injuries, or auto accidents.[3] In this

study focuses on musculoskeletal disorders

Between 12% and 71% of persons worldwide suffer from neck pain, with a considerable fraction having persistent or recurrent symptoms. This issue can be converted to musculoskeletal disorders (MSD). Musculoskeletal disorder (MSD) are prevalent health conditions that affect people of all ages and have a detrimental influence on their quality of life. In addition to their human consequences, these illnesses are known to impose an economic price due to the loss of workforce productivity during the healthcare process.[4] The disease can interfere with everyday tasks, employment, and social interactions, which lowers productivity and raises medical expenses.[5] Neck issues are one of the most frequent MSD complaints, both in terms of discomfort and the detrimental impact they have on quality of life. As technology evolved, it has become more normal for people to sit for long periods of time while using computers and mobile devices. This inactive lifestyle can eventually cause postural problems in the neck, muscular imbalances, and chronic discomfort, lowering overall quality of life.[6]

In the economic perspective of neck pain all around the world, with an estimated cost of 134.5 billion\$, the most costly medical conditions in the US in 2016 were those affecting the neck and lower back. More than 9% of total national medical costs were related to this health problem[7]

Therefore, Around 750 million\$ is the total market volume when considering the size of the global market, particularly for cervical collars.[8] Then in Türkiye, according to TÜİK statistics According to 2023 healthcare spending numbers, medical supply costs totaled 256 billion TL, or 20% of overall healthcare spending. According to data from 2022, among people aged 15 and older, neck-related issues ranked as the second most frequent cause of medical consultations between 2016 and 2022. The second most common health concern causing medical visits for the population aged 15 and over during this time was neck region issues.[9] [10]

In this study, the target group decided for office workers. The underlying reason that there is no specific researches or design development for office workers. Even though there are numerous and interesting researches on ergonomic analysis of office work life. To clarify this, it can be explained as follows:

The proposed study is aimed at office workers, specifically those who sit for long periods of time and frequently interact with visual display terminals. Research reveals that muscle activity and discomfort levels rise over time in such individuals, emphasizing the potential benefits of a potential device designed to minimize these aspects. [11] Furthermore, the neck is the most problematic region for office workers, with higher degrees of discomfort compared to other body parts. [12][13] Prolonged bad posture and physical inactivity are contributing factors to the rising incidence of musculoskeletal disorders and neck pain, especially among office workers. Such pain has a long-term detrimental effect on people's life, impairing both their day-to-day activities and their ability to perform at work. As consequently, neck pain is a prominent cause of disability among office workers, underlining the critical need for targeted preventive and treatment interventions. According to research, once neck discomfort appears, it frequently becomes recurring or persistent, with up to 85% of people reporting a recurrence within a year. [14] This high recurrence rate not only raises healthcare costs but also exacerbates disability, often forcing people to change their job or home surroundings. Neck pain is predicted to become more common as the world's population ages, underlining the importance of effective prevention and therapy techniques. [15] According to projections, 32.5% more office workers will experience neck pain during the next 30 years. [16] As a result, individuals suffering from neck pain eventually become a burden on the healthcare system.

1.1.1 Cervical Neck Supportive Systems in Various Fields

Before begin to this part, the main motivation for developing a neck assistive device on office workers, % 82 of them get musculoskeletal disorders, 78% of the workers don't have regular exercise habit and 66% of them identifies them as less active or inactive.[17]Therefore the other health issue is Drop Head Syndrome (DHS). The people who suffer from this disease cannot activate their neck muscles and their head falls forward uncontrollably due to a significant weakening of the neck extensor muscles (also known as the "chin-on-chest" posture). So that in this case, an ergonomic assistive should be encourage the workers healthcare. Therefore it pursue the research interest on assistive type and designs. After the researches, it can be say that there are 6 main types of assistive neck systems design. To categorize these methods:

BCC: Bracing Collar concept;

HCV: Hooded Collar Vest;

AC: Attachment Concept;

ICC: Inflatable Cervical Collar;

DNC: Dynamic Neck Braces;

RNS: Robotic Neck Systems;

So, that in this part, recent researches and applications examined in this categories. When we look the current researches and ergonomic applications. In BCC type the most com-mon and well known types are Aspen and Philadelphia collar types. [18][19] These are the improved versions of the rigid collars and mainly aim to stabilize the neck posture. When dependable motion limitation is needed without sacrificing comfort for extended wear times, the Aspen collar is frequently used post-operatively, following trauma, or for moderate to severe cervical spine injuries. In contrast, the Philadelphia collar is a traditional hard cervical orthosis made of rigid plastic reinforcements encased in molded polyethylene foam. By limiting flexion, extension, lateral bending, and rotation, it provides the

cervical spine with strong immobilization. Then, Cervigard type collar A Cervigard cervical collar is a semi-rigid neck brace used to immobilize and support the cervical spine. It provides more stability than a soft foam collar but is less rigid than a hard (Philadelphia-type) collar.[20] Then, there is Headmaster collar designed for DHS patients. is a low-profile, lightweight cervical orthosis made especially to help individuals with weak neck extensors control their heads. Instead of limiting full cervical motion, the Headmaster mainly maintains the head in an upright position, in contrast to rigid immobilization collars.[21] To give some alternative example, a study looked into the effectiveness of a personalized neck collar with negative sensory input in treating torticollis, which has not responded to conservative or surgical treatment in the past.[22]

To continue with the second type which is hooded collar vest type, to explain it shortly, A foamy jacket with integrated active and intelligent material-based support components is used in this protective idea. In all flexion, extension, and lateral postures, the smart/active materials can generate opposing forces when placed into embedded pockets through-out the hood's circumference. To exemplify this type, For people with significant neck extensor weakness, the custom 3D-printed cervical collar created for DHS patients is a customized orthotic device intended to improve everyday function and restore appropriate head posture.[23] Then when we look the Halo vests, which are used to treat chronic cervical spine issues, are made up of a metallic ring that is screwed to the patient's skull and is connected to a plastic thoracic cast by metallic rods (struts) that hold the patient's head in place.[24] Then, recently developed parallel mechanism offers dynamic, anatomy-based wearable cervical collar designed to help people with minor neck pain maintain their mobility while decreasing muscular tension. Unlike traditional collars that immobilize the wearer rigidly, this assistive device permits the wearer to conduct about 65% of their natural neck motions as a vest type.[25] Lastly, To help people with neck extensor muscle weakness, particularly in diseases like Amyotrophic Lateral Sclerosis (ALS) and other

motor neuron disorders, the Sheffield Support Snood is a novel, modular cervical orthosis that was co-developed with patients. Its adjustable, detachable supports can be set to restrict particular planes of motion, and it was designed to balance support and mobility.[26] Then, when we continue with the third concept, attachment concept, it is mainly the alternative designs that attached to thing is in the environment mostly walls and chairs to support neck. To exemplify the studies on this concept, firstly, low-cost elastic head support for people with weak neck extensor muscles, especially those suffering from ALS. By attaching a neoprene strap to the user's belt or slacks and using Velcro to secure it to a baseball cap, the gadget forms a flexible "Y" shape support that gently raises the head while permitting natural transverse rotation.[27] Secondly, In order to relieve neck discomfort brought on by extended static neck flexion, such as that experienced during dental procedures or office work, researchers created a lightweight, wearable dynamic neck support device. In order to provide an upward counterforce to the forehead during neck flexion, the device has a band attached to the head that is connected via a force-generating mechanism that is fixed to the upper torso. Without limiting movement, this design converts the static stress on the neck muscles into pressure sustained by the upper body.[28] For the fourth type, Inflatable Cervical Collar, aims to support the neck with an inflatable materials. With this it aims to decrease the pressure on the muscles and try to increase level of comfort on the user. To be able to do that, the most of the studies comes out from the soft robotics are to exemplify, a study offers a passive, pliable neck brace made especially for people with DHS. The multidirectional elastic system of this inventive collar, which is made up of flexible rods and adjustable stiffness bars, allows for unfettered rotation while offering support during forward and backward flexion.[29] Another study demonstrates that a low-profile internal fixation system for cervical support is to stabilize the upper cervical spine. By using soft robotic actuators to stabilize the cervical spine and encourage good posture, this cervical support device is an ergonomic way to lessen neck

pain.[30]

Then to move on with fifth concept which is dynamic neck braces, their usage areas are very wide. But to summarize neck braces are similar to attachment types but generally they offer more Range of Motion (RoM) capabilities compared to other supportive neck de-vices. Then, they generally mounted or supported from the back side of the user. To give some examples from the recent studies, The paper presents a dynamic neck brace based on a parallel manipulator that combines active traction and static support for cervical rehabilitation. To provide controlled symmetric or asymmetric traction, the wearable de-vice uses a 3RPS parallel-robot setup with three linear actuators to guide and assist head movements.[31] Then, A new passive neck orthosis for individuals with Dropped Head Syndrome (DHS) is presented in this study. The apparatus supports the neck during sagittal plane flexion and extension while permitting unrestricted head rotation with a multidirectional, variable-stiffness compliant mechanism. Its main invention is an elastic system that can be adjusted in stiffness using flexible rods, which helps raise the head without overtaxing the flexor muscles of the neck. Unwanted forces are decoupled via a special sliding rail and universal joint system, preserving comfort and normal head mobility. The brace's ability to lessen muscle strain while maintaining necessary neck.[32] Then, for the next one, In order to help DHS patients reach their full natural range of head rotation while maintaining their safety, this research presents a revolutionary design for a dynamic neck rehabilitation robot. The suggested robot is powered by cables, and this is the first neck rehabilitation robot to use a cable transmission. Degrees of Freedom (DOF) are provided by the proposed cable-driven neck rehabilitation robot (CarNeck), each of which is managed by a different 2N type cable-driven joint module.[33] Lastly, created and assessed a new passive neck orthosis specifically designed for people with increasing muscle degeneration. This wearable device, which is made of a compliant multi-directional elastic system that integrates flexible rods and adjustable stiffness components, allows the head

to move freely while providing customized support during sagittal-plane flexion and extension. The orthosis demonstrated its capacity to reduce muscle tension while preserving necessary mobility by successfully unloading the neck extensor muscles without overtaxing antagonists.

Overall, the study identifies a viable ergonomic approach to treating neck extensor weakness that, thanks to its passive, adaptable form, provides both flexibility and supporting help.[34]

For the last concept which is robotic neck systems, which are the systems that developed especially special target group, DHS patients. For people with DHS, particularly those with ALS, this study introduces a unique dynamic neck brace that measures and eventually helps with head and neck movements. This brace, in contrast to static neck collars, simulates the linked rotation and translation of the human head with respect to the trunk using a 3-DOF parallel mechanism (3-RRS: Revolute-Revolute-Spherical joints (RRS)).[35]

In this work, a 6-DOF head and neck exoskeleton with surface electromyography (sEMG)-based control and a parallel manipulator structure is presented. Helping users with weak neck muscles move their necks is the aim. Through the processing of sEMG signals, the system determines the neck muscle force in real time. This data is then sent into an inverse dynamics controller, which produces the necessary assistive force. To lessen undesired contact forces and moments between the user's head and the robotic manipulator, an interaction controller is incorporated.[36]

A closed-chain, parallel robotic neck orthosis designed to aid in the rehabilitation of individuals with cervical musculoskeletal diseases is presented in this study. During rehabilitation activities, the 4-DOF tiny parallel robot is intended to provide gentle support and resistance while actively supporting and guiding neck motions in various directions. The system's closed-loop control architecture and actuators enable safe, repeatable head and

neck motion trajectories, enhancing therapeutic movement precision.[37]

In order to help patients with DHS, particularly those with ALS, hold their heads up and regain voluntary head mobility, the Columbia Exo is an early powered neck exoskeleton. It replicates the natural coupling of head rotation and modest translation with respect to the neck by using a 3-DOF parallel linkage. It can assist in raising the head and enabling the wearer to actively change head position while reducing neck muscular strain when operated via a straightforward user interface (similar to a joystick). An enhanced version of this concept is the Utah Exo. The 3-RRS parallel mechanism remains the same, but stronger linkages, improved structural stability, and a more accurate joint design are added to better deliver torque and prevent undesired motion or vibrations. By lowering slack and misalignment, these enhancements increase the device's responsiveness and safety. While enabling users to maintain comfortable, natural mobility rather than rigid fixation, the Utah Exo outperformed the original Columbia Exo in testing by supporting head posture, reducing muscular load, and facilitating smoother, more precise head tracking.[38]

1.1.2 Mathematical Model of Cervical Spine and ADAMS

So, when we look at the mathematical roots of the neck, Numerous mathematical models have been created to depict the intricate biomechanics of the human cervical spine. Huston et al. developed a mathematical model with nine rigid bodies, each with six degrees of freedom, to represent ligaments, muscles, and discs with springs and dampers.[39] Then, Researchers constructed a three-dimensional mathematical frame-work including nine lumped masses to study head motion during high-acceleration rear and side impacts.[40] More realistic models have been brought into the literature by include muscles. A model with 15 muscle pairs with nonlinear constitutive relationships was developed and compared to previous models under various stress circumstances.[41] Then,

Williams and Belytschko incorporated 22 muscle types and modeled ligaments and intervertebral discs as nonlinear springs.[42] Then during the muscle analysis Hill-type model starts to apply on the simulations. A phenomenological model that depicts the mechanical behavior of skeletal muscle during contraction is the Hill-type muscle model. Based on his studies examining the connection between muscle force, length, and contraction velocity, A.V. Hill initially identified it in 1938.[43] While yet being mathematically and computationally tractable, this structural model well represents key muscle properties as length-dependent force, velocity-dependent force output, and elastic energy storage. It serves as the foundation for the majority of musculoskeletal simulations (OpenSim, AnyBody, etc.) and is extensively utilized in neural control, biomechanics, rehabilitation, and robotic systems. Then for the tendon dynamics, Three musculotendon models—equilibrium, damped equilibrium, and rigid-tendon—used in simulations of human and animal movement are compared in this research based on their computing speed and biological correctness. Particularly at low activation levels, the researchers discovered that the damped equilibrium model, which incorporates light damping to prevent numerical singularities, generated force profiles that were comparable to the equilibrium model but simulated noticeably more quickly. The rigid-tendon model was the quickest and best suited for situations involving short tendons, but it was less accurate for muscles with lengthy tendons.[44] Whiplash injuries in rear impacts were examined using a multi-body dynamics model and the Modified Extended Backward Differentiation Formula with Variable Step Sizes (MEBDFV) solver code. The impact of design choices on whiplash phases was also examined.[45] From a different angle, Sendur et al. used an ADAMS model of the cervical spine to assess how different design characteristics affected the four stages of whiplash injury.[46] To continue with the development of cervical spine models A 48-degree-of-freedom nonlinear musculoskeletal model was then created, emphasizing kinematic reactions and passive muscle characteristics.[47] Adding hyoid muscles to a

musculoskeletal model improved its accuracy in OpenSim.[48]The OpenSim model and studies on human subjects have been used to investigate the role of soft tissue in the human cervical spine system.[49]

Subsequently, to continue with the biomedical applications in ADAMS software is very rare. Firstly, In order to replicate the intricate biomechanics of a golf swing, the researchers in this work used the ADAMS Multibody Dynamics (MBD) software in conjunction. In order to enable thorough kinematic and kinetic analysis, the full-body model was built to accurately depict the golfer's skeletal and joint structures with several degrees of free-dom. The researchers utilized ADAMS to replicate the full swing motion, examine joint angles, ground reaction forces, and the club's dynamic behavior by importing 3D motion capture data from a proficient golfer. In order to show that ADAMS can accurately simulate the complex interactions between the golfer's body and the equipment during a high-speed, multi-joint movement, the simulation results were compared to experimental measurements.[50]For another example hip's region analyze in a study that the interaction between body segments and a hip prosthesis was the primary emphasis of this study's multibody dynamics (MBD) model of human walking, which was created using ADAMS[51]

1.1.3 Experiments on Office Workers and Neck Pain

When we look at the experiments on the neck pain and office workers there are two types of researches. First, the ergonomic approach looks at the risk factors that exist in the workplace, analyzes the particular tasks and activities that are done there, assesses the impact of neck pain on work performance and overall productivity, and evaluates the efficacy of training programs. To exemplify, a research emphasizes how low task diversity, repetitive static postures, and poor workplace ergonomics all lead to long-term muscle tension and discomfort. The

writers stress the significance of identifying modifiable physical risk factors to guide targeted ergonomic interventions and preventative efforts to reduce work-related neck pain and its impact on productivity, given the high prevalence and recurrence of neck discomfort among employees in offices.[52] During a full day of computer work, office workers with and without neck-shoulder pain had their neck and shoulder postures compared in another field study.[53]

The second type is that about detecting the muscle activity during specific task using sensors. Common sensors using in this research are EMG, IMU sensors and movement analysis program with using the cameras. The second approach uses wearable sensors to identify muscle activity during particular tasks. Inertial measurement unit (IMU) and surface electromyography (EMG) are often utilized sensors in this kind of study. Real-time data on muscle activation levels and exhaustion during different activities is provided by surface EMG sensors, which monitor the electrical activity generated by skeletal muscles.[54] In contrast, IMU sensors usually integrate gyroscopes, accelerometers, and occasionally magnetometers to record precise motion information like posture, joint angles, and dynamic head or neck movements.[55] Researchers can thoroughly examine the connection between muscle activity and motion patterns while doing office work or utilizing neck-supporting devices by integrating EMG and IMU data. For example study, investigated the physical risk factors that contribute to office workers' non-specific neck pain. The scientists found important risk variables by analyzing 20 research, including little task variation, self-reported excessive muscular tension, and poor workstation ergonomics (such as a keyboard that is too close to the body). Despite being a review, this study emphasizes how contemporary research commonly uses IMUs to record posture and movement data during computer-based tasks and EMG to track neck muscle activation levels, providing objective measurements of head/neck position and muscle

strain. When creating ergonomic therapies, researchers may establish a connection between physical risk factors and measurable muscle loads and motion patterns thanks to this combined sensor approach.[56]To give some more examples between the EMG and neck muscle activities. The study contrasted female office workers with persistent neck muscle pain (trapezius myalgia) to healthy controls to examine the association between torque, muscle activity, and contraction speed. The researchers used surface EMG and isokinetic dynamometry to find that workers with trapezius myalgia had reduced activation of the trapezius muscle, particularly during slow concentric and eccentric contractions, and lower shoulder abduction strength, while the pain-free deltoid muscle was unchanged. These results demonstrate that whereas synergistic muscles are not greatly affected, persistent neck pain specifically impairs the function of the afflicted muscle.[57]To continue with the analyzing with the sample office tasks, using surface EMG, the study examined the patterns of neck and shoulder muscle activation in office workers with and without symptoms after they worked at a keyboard for 60 minutes straight. In contrast to asymptomatic workers, workers with neck and shoulder pain demonstrated increased upper trapezius muscular activity, more sustained low-level muscle loading, and decreased muscle rest. This finding emphasizes how extended repetitive tasks might raise the risk of muscle fatigue in susceptible groups.[58]Therefore, using motion analysis, a follow-up study looked at the postures and movements of the head, neck, and shoulders while doing the identical keyboard job. The development or persistence of neck and shoulder diseases in office environments may be attributed to limited movement variability, as symptomatic office workers maintained more prolonged static head and neck postures with less frequent posture variation than asymptomatic workers.[59]Recently, similar tasks doing applied on smartphones.[60]Moreover, The researchers discovered that those with persistent neck discomfort exhibited decreased cervical spine mobility, changed movement patterns,

and greater muscle co-contraction, particularly in the upper trapezius and cervical erector spinae, using motion capture and surface EMG. These results imply that ineffective muscle use and restricted spinal movement are linked to persistent neck pain, which may lead to ongoing discomfort and dysfunction when performing daily tasks.[61] This study examined the posture of the head, neck, and upper back when using a desktop, laptop, and tablet computer while standing and sitting. In order to determine how various devices impact muscle loading, the researchers evaluated postural angles as well as muscle activity in the neck and shoulder region using motion analysis and surface EMG.[62] Then, in another example EMG is used to measure the activation of the upper trapezius and neck extensor muscles in healthy office workers as they complete standardized computer and desk duties.[63]

For the tracking researches, earlier studies, Using a three-dimensional motion capture system with numerous cameras to precisely measure neck movements in flexion, extension, lateral bending, and axial rotation, the study examined the Active Range of Motion (AROM) of the head and cervical spine in young individuals in good health.[64] Then, a lab investigation, 18 individuals' head flexion angles were recorded while they were seated and standing while performing three frequent smartphone tasks: texting, web browsing, and viewing videos. Through motion analysis, the researchers discovered that the average head flexion was between 33° and 45° from vertical, that the highest forward head angle was produced by text messaging, and that the most flexion happened when seated as opposed to standing.[65] Then, when we look back to research about the postures during the using smartphone. Using (sEMG) to quantify activation of the cervical erector spinae (CES) and upper trapezius (UT), this study investigated the effects of varying neck flexion angles (0° , 15° , 30° , and 45°) during 90 seconds of smartphone use on muscle activity in users with and without chronic neck discomfort.[66] Then, about the fatigue during using smartphones. A study, 14 young people's head flexion angle

and neck muscle fatigue throughout an hour of smartphone use were examined. While seated, participants held cellphones, and surface EMG sensors collected muscle activity while a gyroscope continually measured their head angles. The study's objectives were to assess posture using the Rapid Upper Limb Assessment (RULA) tool and connect increasing head flexion with symptoms of muscle tiredness as determined by EMG median frequency.[67] Another remarkable study, 23 male volunteers were grouped according to the health of their neck muscles using the flexion relaxation ratio (FRR) to assess the effects of stationary versus gradually moving monitors on neck posture and fatigue during a 50-minute document-writing job. In order to measure forward head posture, researchers employed surface EMG to measure cervical erector spinae fatigue using the flexion relaxation phenomenon (FRP) and craniovertebral angle (CVA) from camera images. In comparison to a fixed monitor, the study discovered that the moving monitor decreased the frequency of forward head posture and marginally lessened neck fatigue, and that FRR values decreased following the task, indicating muscular fatigue.[68]

1.2 Research Gap and Aims

While there are many different types of neck orthoses on the market, such as soft collars, stiff collars, and even robotic systems, these solutions are mostly intended for managing pre-existing neck issues or for rehabilitation following an injury. There is currently no workable tool to actively avoid neck pain or maintain good neck posture during extended desk work, despite a wealth of research examining office ergonomics, risk factors, and muscle activity patterns among office workers. Furthermore, no research has yet successfully merge multi-body dynamic modeling insights with an ergonomic product to develop a neck support system that combines supportive and preventive features for daily usage.[69][70]

The goal of this thesis is to close this obvious gap by creating the novel, simulation-informed Neck Assistive Cervical Kit (NACK), an assistive technology intended to improve neck posture, lessen muscular fatigue, and help office workers avoid neck pain.

1.3 Structure of Thesis

Therefore this study aims to develop a preventive neck orthosis that can balance the neck posture decrease the muscle activity to prevent from neck fatigue and naturally neck pain. To be able to that first a detailed neck model which has ligaments, vertebrae, active and passive muscles to set to design constraints for the neck stiffness. Therefore to decide this parameter a design of experiment will applied to reach optimal stiffness value. After the stiffness parameters set the design constraints will examined and the design will be developed with iterations and it will be finalized. After the design constraints determined the CAD design will be drawn then it will be printed in the 3d printer and absolutely, the materials and the type of the printer selected according to design constraints. After getting the first prototype from the printer, the experimental procedure will be planned and performed. The first objective of the experiments to get EMG data to analyze muscle activation. Second objective is to get position and acceleration data from IMU sensor and put them into the detailed neck model to examine the forces on the neck including deep flexor neck muscles.

1.4 Roadmap

The thesis's remaining sections are organized as follows: Chapter 2 describes the detailed neck model and setup of the Design of Experiments (DOE) in the simulation. Furthermore in this section the parameters that using to determine the stiffness will be shown. Chapter 3 unveils the design parameters and design iterations during the

Computer-aided Design (CAD) process and also this chapter includes the 3d manufacturing process. In the Chapter 4 the steps of the experimental procedure shown and the reasons behind the experimental examined and results shown in graphics, tables etc. to compare target objectives and experimental data. Lastly Chapter 5 is about for conclusion and mention about the future work.



2. CERVICAL SPINE MODELLING

The neck model will be thoroughly investigated in this chapter. The model was first created and modified from an OpenSim crash test simulation to be able to do accurate modal and frequency response analysis. The 50th percentile of male anthropometric data served as the basis for defining all model parameters. The head, individual vertebrae C1–C7, and the T1 thoracic vertebra for realistic boundary conditions are all included in the model's cervical spine component. The model is backed by input data gathered from the literature and includes all of the fundamental anatomical components, including muscles, ligaments, and tendons.[44][49][41][71]The model's intricate structure, including its parts, ligaments, and muscle groups, is shown in the sections that follow.

2.1 Neck Model

The model has 9 parts the parts are the C1-C7 bones and T1 and skull which are the parts of the spine, also known as the vertebral column, is a strong, flexible structure that runs vertically along the back. It is composed of a number of separate bones known as vertebrae. The five areas of these vertebrae are lumbar, sacral, coccygeal, thoracic, and cervical. The cervical spine contains seven vertebrae, designated C1 through C7. The skull is connected to C1, and the T1 is connected to C7 and these parts are designed in CAD which can be seen in figure 2.1

Figure 2.1: Neck Model

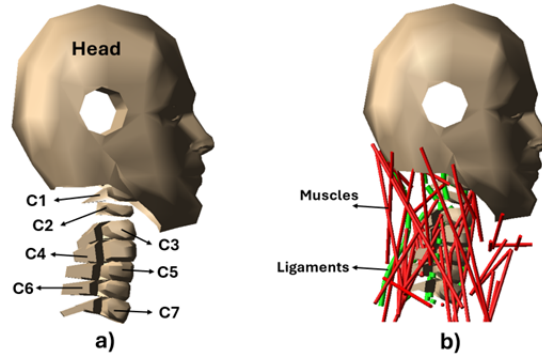


Table 2.1 provides a summary of the model's parameters, including mass, Center of Mass (COM), and moment of inertia. Every piece of inertial and geometric data comes from earlier research.[49]

Table 2.1: Mass, mass moment inertia, and CoM properties of the parts in the model

PartName	Mass (kg)	Ixx (kgm ²)	Iyy (kgm ²)	Izz (kgm ²)	CoM position (x,y,z) (m)
T_1	57.5	1.97	0.37	1.97	(0, 0, 0)
C_7	0.38	2.56E-3	3.17E-3	7.33E-4	(-0.063, 0.419, 0)
C_6	0.22	2.44E-4	1.18E-3	6.48E-4	(-0.059, 0.438, 0)
C_5	0.26	7.33E-4	1.71E-3	5.86E-4	(-0.052, 0.462, 0)
C_4	0.20	1.83E-4	6.84E-4	4.52E-4	(-0.049, 0.475, 0)
C_3	0.15	4.03E-4	5.25E-4	3.18E-4	(-0.048, 0.499, 0)
C_2	0.15	4.03E-4	5.25E-4	3.18E-4	(-0.049, 0.517079, 0)
C_1	0.15	4.03E-4	5.25E-4	3.18E-4	(-0.052, 0.534, 0)
Head	4.12	2.53E-2	1.83E-2	0.027E-2	(-0.032, 0.618, 0)

For visualization, a CAD model of every component is appended to the center of mass. The mobility of the components in relation to one another is defined by joints. There are two dummy components between each vertebra.

Revolute joints are used to link each vertebral and dummy pieces. It is feasible to determine the flexion-extension, lateral, and axial rotation of each spine using this modeling technique. Every vertex's translational degrees of freedom are completely limited. For correlational reasons, the experimental circumstances must be replicated by axial rotation between C1-head, lateral rotation between C1-head, C1-C2, and flexion-extension of C1-C2[49].The model features 20 degrees of freedom (DoFs) in total.

2.1.1 Ligaments

Then, ligaments, one of the most important components that stabilizes and limits mobility in the cervical spine and neck area are ligaments. These strong, fibrous structures give joints support and link vertebrae. The range of motion (ROM) of the joint in all flexion-extension, lateral bending, and axial orientations is also specified by the ligaments.[72]

The model used [49]. The force-length (f^{lig}) and force-velocity (f^v) relationships determine the force produced by the constitutive ligament model. The length of the ligament (l^{lig}) determines the first, while the loading rate (v^{lig}) determines the second. Equation 2.1 provides the ligament's constitutive model:

$$F_{\text{ligament}} = f_{\text{lig}}^L (l^{lig}) f_{\text{lig}}^v (v^{lig}) \quad (2.1)$$

When the ligament is strained, three different stress response areas are activated. As the deformation increases, the stress in the first area, known as the toe region, rises exponentially. As it moves from the first to the second region, the response becomes linear. In the third region, the stiffness decreases nonlinearly until the fiber ruptures. Only under tensile loading do the ligaments produce force. Equation 2.2 is used in the model to compute the ligament strain (ϵ^{lig}). Equation 3 is used to compute the ligament force after the strain has

been established. The ligament force is zero when ϵ^{lig} is negative.

$$\epsilon_{lig} = (l^{lig} - l_0)/l_0 \quad (2.2)$$

Equation 3 defines the toe area as the point at which

$$\epsilon^{lig} < \epsilon_{toe}$$

. Here, the toe modulus (ϵ_{toe}), ligament strain, and ligament cross-sectional area (A^{lig}) are used to parameterize the generated force. The literature provides the values for toe strain and toe region modulus [49], [72], [73]. The cross-sectional area ranges from $1.09 \times 10^5 m^2$ to $11.28 \times 10^5 m^2$, and the toe modulus ranges from 0.392 MPa to 25 MPa.

$$f_{lig}^L(l^{lig}) = \begin{cases} 0, & \epsilon^{lig} < 0 \\ A^{lig} \epsilon^{lig} E_{toe}, & \epsilon^{lig} < \epsilon_{toe} \\ A^{lig} (E_{toe} \epsilon_{toe} + E_{lin}(\epsilon^{lig} - \epsilon_{toe})), & \epsilon^{lig} > \epsilon_{toe} \end{cases} \quad (2.3)$$

Lastly, the loading rate affects the ligament's force as well, particularly at high loading rates. There are two areas in the force–velocity relationship. In the first region, the quasi-static loading rate is 10 mm/s [49], while in the second, the log rating is multiplied by the slope m_{rate} . Equation provides the constitutive relationship between ligament force and ligament velocity.

$$f_{lig}^v(v^{lig}) = \begin{cases} 1, & v^{lig} < 10 \text{ mm/s} \\ 1 + m_{rate} \log\left(\frac{v^{lig}}{10 \text{ mm/s}}\right), & v^{lig} \geq 10 \text{ mm/s} \end{cases} \quad (2.4)$$

2.1.2 Muscle Models

Then after the ligaments for the muscles, muscles in the models modelled according to the approaches on the researches.[74][75]To comprehend how force is produced and how muscles work, one must comprehend the basic anatomy of muscle fibers. This section will explore the complex realm of muscle fiber structure by looking at sarcomeres and muscle cells.

The human body's workhorses, muscles, are composed of specialized cells called muscle fibers. Because of their distinct form and function, these muscle fibers can contract and produce the force needed for a variety of body movements. The connection between force and length is essential to comprehending how muscles produce movement and power in muscle mechanics. Muscles are extremely specialized biological machines, and the length of the muscle at the start of the contraction directly affects how well it contracts. Since it specifies the range in which a muscle can work at its best, this information is essential for maximizing muscular performance. The muscle model that used in this detailed model, in the 1930s, American physiologist Archibald Hill created the Hill-Type Model, a key idea in muscle physiology and biomechanics. This model offers a clear and concise illustration of how muscles contract and produce force. Two key relationships that determine muscle behavior must be understood in order to comprehend the Hill-Type Model:

The force-length relationship describes how the length of a muscle affects the force it can produce. Within a certain range of lengths, muscles function at their best. The muscle's ability to produce force diminishes at lengths that are either shorter or longer than this ideal range. grasp how muscles work in different contexts, like lifting weights, where muscle length affects the amount of force that can be applied, requires a grasp of the force-length connection.

The force-velocity relationship explains how a muscle's capacity to produce force is influenced by the speed at which it contracts. When a muscle contracts isometrically, or without changing length, it can produce its maximum force; however, when the contraction speed increases (concentric contraction) or lowers (eccentric contraction), the force that the muscle can produce diminishes. This link has significant ramifications for sports like sprinting, where muscle contraction speed is a key component of success. Since it provides the most advanced muscle modeling methodology, the Hill-type modeling approach from the literature is used to depict active and passive muscles. The length, velocity, and activation level of the muscle fibers—which are determined by brain excitation—all affect the muscular forces. Experimental data from the literature has been used to validate this model [44].

For the active muscles, the suggested model makes advantage of first-order activation dynamics[44].

$$\dot{a} = \frac{u - \hat{a}}{\tau} \quad (2.5)$$

More precisely, Equation 2.5 illustrates how a first-order differential equation is used to determine the muscle activation (a) from the muscle excitation (u). The neuronal activity, represented by the parameter u in this equation, has a value of 1 [44].

$$\hat{a} = \frac{a - a_{\min}}{1 - a_{\min}} \quad (2.6)$$

$a_{\min}$ is the lowest amount of muscle activation that is allowed in Equation 2.6. in order to avoid numerical singularities. According to literature, $a_{\min}$ is set to 0.01.[44]

$$\tau = \begin{cases} \tau_A(0.5 + 1.5\hat{a}) & \text{if } u > \hat{a} \\ \frac{\tau_D}{0.5 + 1.5\hat{a}} & \text{otherwise} \end{cases} \quad (2.7)$$

The activation and deactivation time constants in Equation 2.7 are (τ_A) (0.01 ms) and (τ_D) (0.04 ms), respectively [49]. Calculating the muscle-tendon length, which is provided by Equation 2.8, is the first step in figuring out the force in the muscle.

$$l^{\text{MT}} = l^M \cos(\alpha) + l^T \quad (2.8)$$

where l^T is the length of the tendon, l^M is the length of the muscle fiber, and l^{MT} is the muscle-tendon length. l^{MT} normalizes the input displacement because the muscle force depends on the normalized fiber length. Following the determination of l^M , the normalized muscle force is computed using Equation 2.9 and the graphs in Figure 2.3

$$F_{\text{muscle}} = f_0^M \left(a f^L(\hat{l}^M) f^v(\tilde{v}^M) + f^{PE}(\hat{l}^M) \right) \quad (2.9)$$

The muscle force equation 2.9 represents a Hill-type model that combines active and passive force components. The total force F_{muscle} is calculated as the product of the maximum isometric force f_0^M with the sum of active and passive contributions. The active component consists of three factors: the muscle activation level $a \in [0, 1]$, the force-length relationship $f^L(\hat{l}^M)$ (where $\hat{l}^M = l^M/l_0^M$ is normalized fiber length), and the force-velocity relationship $f^v(\tilde{v}^M)$ (with $\tilde{v}^M = v^M/v_{\text{max}}^M$ being normalized velocity). The passive elastic force $f^{PE}(\hat{l}^M)$ accounts for the tissue's intrinsic stiffness. This formulation captures the fundamental length- and velocity-dependent properties of muscle contraction, where the active force generation is modulated by both the instantaneous muscle geometry ($\hat{l}^M$) and contraction dynamics ($\tilde{v}^M$), while the passive component becomes significant primarily at extreme lengths.[44]

2.2 Improvements of the Spine Model

The model's biomechanical framework, which Heidari et al. and others moved from OpenSim to ADAMS, was explained in the preceding sections.[76] The objective of this study is to finish the neck model's kinematic analysis. This was accomplished by applying motion to the joints in ADAMS along the x, y, and z axes and limiting the cervical vertebral movements between C1 and C7 in order to meet the kinematic analysis's 0-DOF (degree of freedom) requirement. Since sine and cosine functions are basic periodic functions for oscillatory motion, they can be used to model the wave motion of head-neck flexion and extension where the sine function can be seen in the equation below.

The sinusoidal function is given by:

$$\Theta(t) = A \sin(\omega t + \phi) \quad (2.10)$$

Three essential factors in a sinusoidal model can be used to characterize the motion of head-neck flexion and extension. The maximum angle of movement, such as $\pm 30^\circ$ is represented by the amplitude (A), which shows the maximum range of flexion (forward) and extension (backward). The oscillatory motion's speed is determined by the angular frequency (ω), which may be computed as:

$$\omega = 2\pi f \quad (2.11)$$

where f is the frequency in Hertz (Hz). The speed at which the motion alternates between flexion and extension is determined by this parameter.

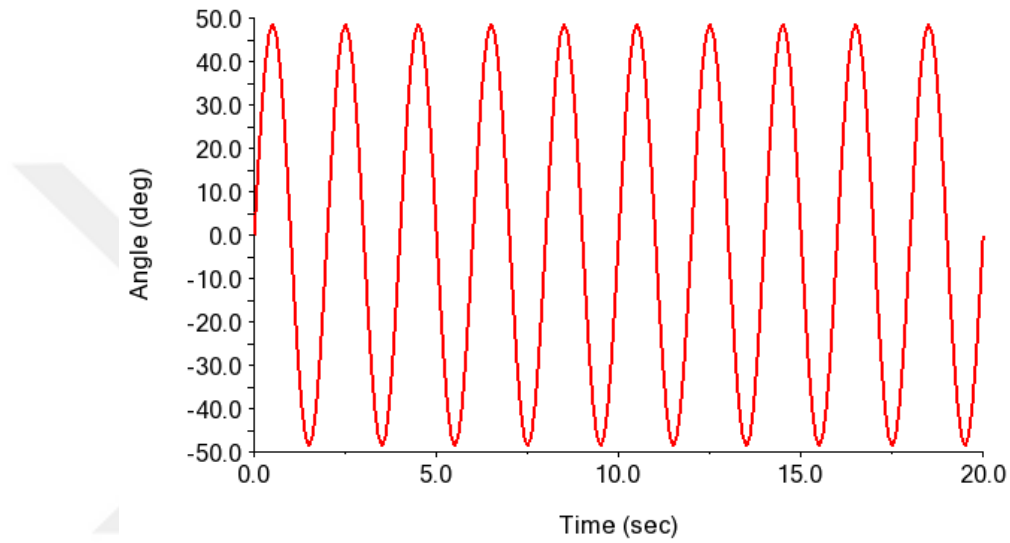
Lastly, the phase shift (ϕ) modifies the motion's initial position, enabling the model to start at neutral position. In this analysis, the aim was to reach maximum neck flexion

which is 48.5° (approximately 0.85 radian). The resulting function is:

$$\theta(t) = 0.85 \sin(\pi t) \quad (2.12)$$

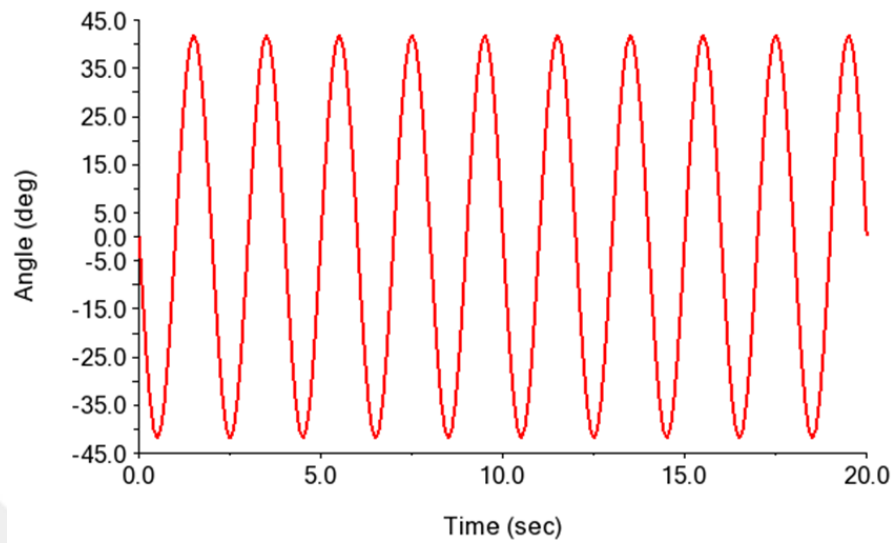
with frequency of 0.5Hz and 10 revolution in 45s

Figure 2.2: *Flexion-Extension*



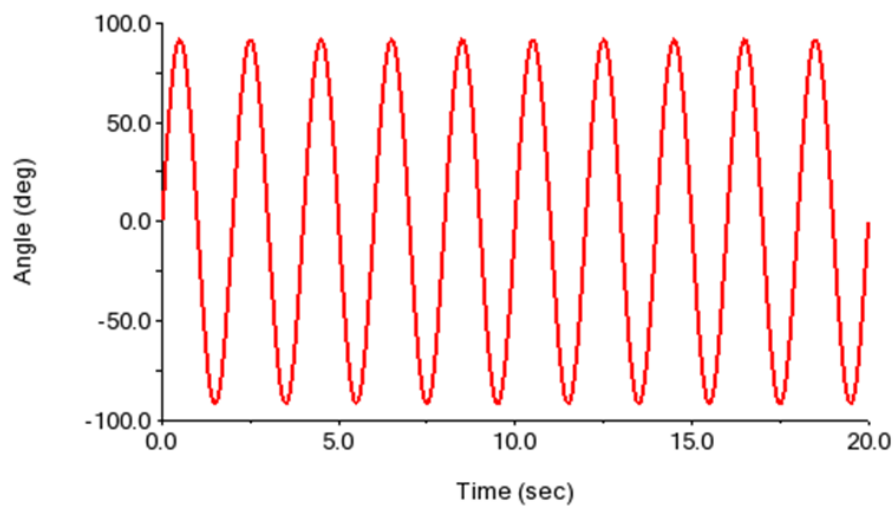
Therefore for lateral neck flexion the aim was to reach 42° which is the maximum of lateral neck flexion according to researches.[77] $0.733 \cdot \sin(\pi \cdot \text{time})$ is the function that reach to 42 degrees for lateral flexion. Lastly , frequency is same with flexion extension and naturally, the total revolution is same.

Figure 2.3: *Lateral Flexion*



Lastly, for the rotation 180 degrees of range of motion aimed. Then to be able reach that level the function applied like this: $1.6 \cdot \sin(\pi \cdot \text{time})$ is the function that reach to 91 degrees for rotation. Then the frequency is 0.5 Hertz (Hz) and revolution is 10 in 20 seconds.

Figure 2.4: *Rotation*



To conclude this part, the head, T1 segment, and vertebrae C1–C7 are all included

in the model, giving it a total of 20 degrees of freedom (DOF). Each component's mass, center of mass, and moments of inertia were determined using information from the literature. With two intermediate (dummy) elements positioned in between each pair of vertebrae, joints were employed to depict the relative motions between the vertebrae. Revolute joints, which connect the vertebrae and the dummy components, enable the simulation of flexion-extension, lateral bending, and axial rotation for every vertebral level. The model fully constrains all translational degrees of freedom between vertebrae.

Axial rotation between C1 and the head, lateral bending between C1 and the head, and flexion-extension between C1 and C2 are all restricted to maintain conformity with experimental conditions. Using ADAMS software, this biomechanical model was created to run intricate multi-body dynamic simulations. This model's parametric structure is one of its main advantages; it enables variables to be changed for various patient groups or damage scenarios, enabling the creation of personalized orthotic solutions.

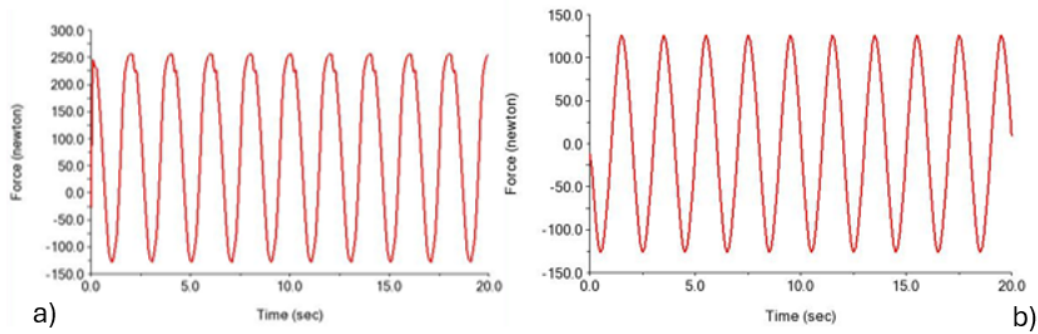
3. DOE AND PROTOTYPE DESIGN

This chapter will explore the design, production, and DoE studies conducted using the knowledge obtained from the comprehensive neck model. After finishing the model validation in ADAMS, a design scenario was established, and the design was created in a CAD environment using the acquired data. Subsequent to that, static assessments were conducted utilizing Finite Element Analysis (FEA) methods, and ultimately, the prototype was developed employing 3D printing techniques.

3.1 Motion Validation on Model

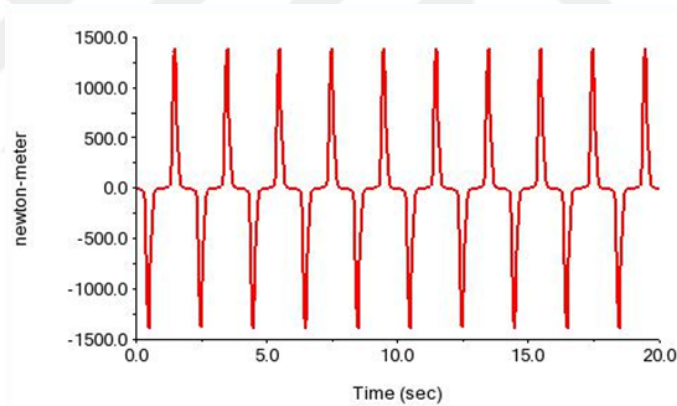
In this section, the process of the design examined and to get this design the DoE process of it deeply explained. In the previous chapter the kinematics of the neck model has done and the neck can be moved according to the given signal. After that, to check the spring mechanism the springs located different places on the neck model. Therefore an error occurs. In ADAMS when the movement specified in the motion block the springs become passive and it unable the measure their effects on ROM. So to solve this, the motions analyze in the postprocessor and export the force data to use on the springs. To be able to do this the force data imported simulation again and given as a force application on the head In the figure 3.1 it can be seen flexion extension and lateral flexion

Figure 3.1: a): Force on flexion-extension b): Force on lateral flexion



Naturally, for the rotation the torque on the y -axis taken for the data which is can be seen in the figure 3.2.

Figure 3.2: Torque during rotation



As a result, due to lack of capability of the motion in ADAMS the motion data have to transfer to the force function for flexion-extension and torque function for rotation. To continue with other task is to combine to these 3 movements in one simulation.

3.1.1 Combining the Neck Movements

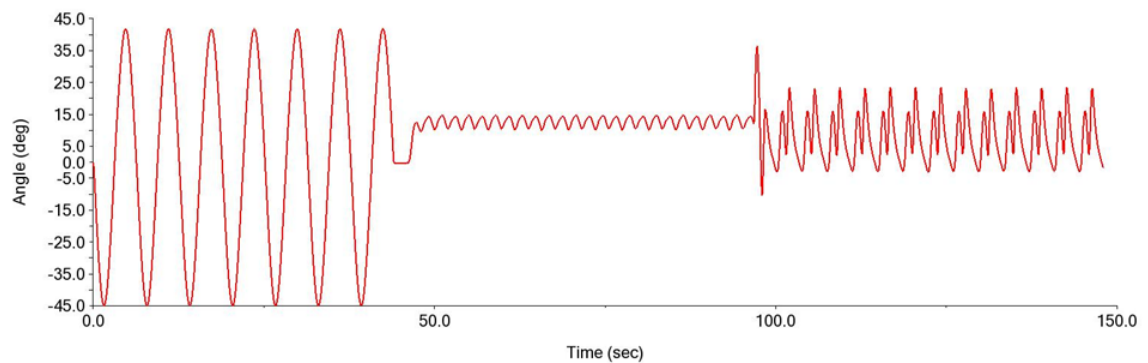
As it mentioned in the previous part the force function applied on the applied on Adams the forces and torques shown on the previous parts became data parameter for the

function. The function for the flexion extension can be seen at the below.

$$f(\text{time}) = \begin{cases} \text{AKISPL}(\text{time}, 0, \text{HumanNeck.A}, 0) & \text{if } \text{time} - 44 \leq 0 \\ 0 & \text{otherwise} \end{cases} \quad (3.1)$$

is used to define a conditional input for the neck system. This command means that until the simulation reaches 44 seconds, force parameter holds a constant value of 23. Since when the body leftover with just the dummy weights it drops like a empty body. Therefore constant 23N is for the cocantracted neck body. Then for data, in this case it represents the force data taken from flexion-extension in another way, the force applied on along to x-axis. The input changes to a time-varying function specified by an Akima spline interpolation that nicely fits the data once the time exceeds 44 seconds. HumanNeck. A dataset over time. The Akima spline method is a piecewise polynomial interpolation approach that minimizes sharp slope changes between segments while producing a smooth curve through a given collection of data points. The Akima approach is more suitable for modeling biological motion or force profiles that need natural, continuous transitions free of oscillations than typical cubic splines because it is more resilient to abrupt changes or outliers in the data. By using this method, the neck model can precisely control when the dynamic motion begins during the simulation and reproduce realistic time-dependent behavior[78]. In the figure 3.3 below, the RoM of the movement can be seen.

Figure 3.3: *Range of motion of flexion-extension*

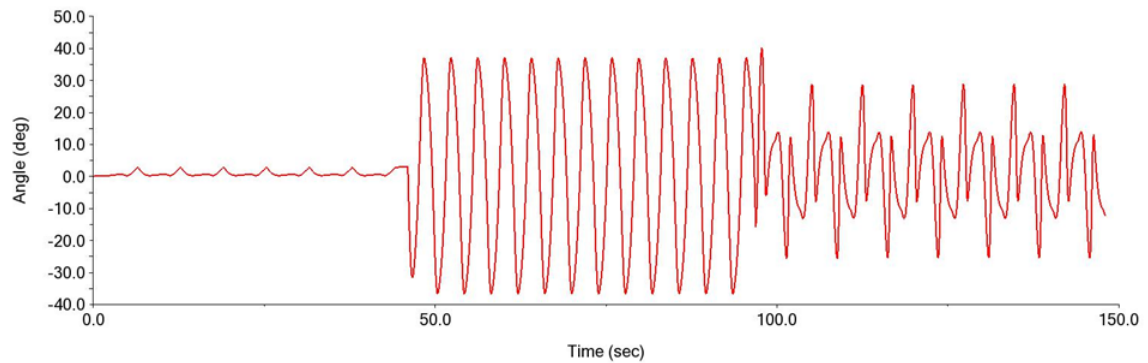


Then, when we continue with the lateral flexion the function almost same with the flexion-extension. Since just the force applied on the axis changed which is become to along z-axis.

$$\begin{aligned} & \text{IF}(\text{time} - 45 : \text{AKISPL}(\text{time}, 0, \text{.HumanNeck.B}, 0), 0, 0) \\ & + \text{IF}(\text{time} - 96 : \text{AKISPL}(\text{time}, 0, -\text{.HumanNeck.B}, 0), 0, 0) \end{aligned} \quad (3.2)$$

In the function the name of the data set changed due to the difference with the first one, furthermore the initial cocontracted force is not necessary for the lateral movement. The only change is during the flexion extension the system should be stay inactive. Therefore it should be stay in active during the rotation. According to this necessity the function designed. The RoM can be seen in the figure 3.4 below.

Figure 3.4: *Range of motion of lateral flexion*

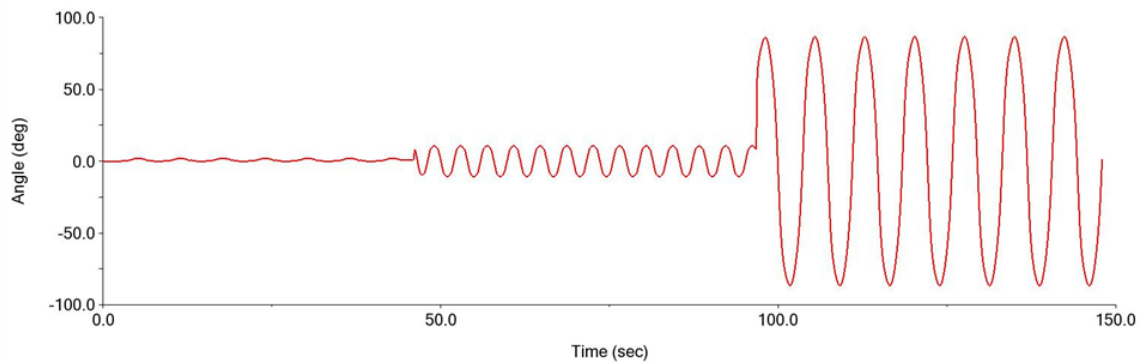


Lastly for the rotation the torque applied on the point center of the mass of the head and around the y-axis of it. Here is the function of the rotation.

$$\text{IF}(\text{time} - 96.76 : 0, 0, \text{AKISPL}(\text{time}, 0, \text{.HumanNeck.C}, 0)) \quad (3.3)$$

In this function the function get activated after the lateral movement finished. Furthermore, the name of the data set changes due to the movement change. The RoM of rotation can be seen in the figure 3.5 below.

Figure 3.5: *Range of motion of rotation*



To conclude this section, After the issues solved in the transformation of motion to force, different types of movements can be applied and combined on the same simulation run.

3.2 Design Approach

In this part the reasons behind the selected design are discussed. An outline of the design iterations applied to the head component is shown in this part, which also discusses the reasoning behind the chosen design. As stated in the beginning, this study assessed six distinct neck support techniques, including lifting help, posture correction, rehabilitation, and general support. The ergonomic compatibility and vibration control effectiveness of these techniques were the main criteria used for evaluation. Attachment Concept (AC) was chosen for the final design since it was determined to be the most effective strategy for accomplishing the study's objectives based on this thorough evaluation.

The following briefly describes the rationale behind choosing the Attachment Concept (AC) as the best approach: According to ergonomic theory, office workers typically experience more discomfort from continuous usage of support designs that come into close contact with their necks, such as classic collars, vest-style supports, or inflatable collars. Furthermore, it is challenging to develop a design that can adequately support the neck in

all degrees of freedom (DoFs), with the exception of collar type. As a result, early in the design process, ideas that were inflatable or vest-like were disregarded.

Even while robotic and dynamic neck braces can support all degrees of freedom, their sophisticated components such as linear actuators, sensors, and control units—make them costly, unwieldy, and unsuitable for daily usage in an office environment. Furthermore, Dropped Head Syndrome (DHS) patients, who need substantial external assistance because of severe muscle weakness, are the main target audience for these devices. In that situation, the additional weight of these robotic systems is usually tolerable, but for healthy office workers, it would be inconvenient and uncomfortable.

The goal of this endeavor is preventive rather than rehabilitative when comparing the collar-type and attachment-type ideas. Instead than immobilizing or treating an existing disease, the purpose of using springs is to store and release energy to support the neck muscles and lessen tension. We plan to steer clear of collar-type designs since they are more closely linked to medical rehabilitation equipment. Rather than being marketed as a medical device, the attachment approach allows for a more efficient integration of the spring mechanism and positions the device as an ergonomic accessory, such as one that is attached to an office chair.

As a result, the Attachment Concept was chosen as the most useful, approachable, and successful strategy for accomplishing the NACK project's preventive objectives.

3.2.1 Design Iterations

After selected the design type. A scenario made up for the getting the ideal design. 200N force will apply on the head since it is almost the strength can a male neck produced.[79]Then springs set to 150N/mm to understand their behavior. With that the iterations more focus on the location of the springs.

During the analysis phase, different spring configurations and positions were tested in

order to determine the best arrangement for supporting the neck. In addition to providing stability and functionality when firmly fixed to an office chair, the springs' design permits simple attachment and detachment to enable maintenance or user-specific modifications. Furthermore, the design includes enough room for appropriate spring deflection, which enables the system to support normal head and neck movements while still offering the desired level of support.

The main goal of this design phase was to incorporate the spring mechanism into the neck model in order to ascertain the best location for it within the available workspace. In order to assess how elements like the quantity of springs will impact overall performance and user comfort, this procedure involved examining several configurations. The objective was to properly position the springs without sacrificing the model's ergonomic acceptability or structural soundness. This phase was built upon the dynamic neck model created in ADAMS, which directed the design iterations and concept development by offering crucial insights into motion behavior, load distribution, and realistic biomechanical restrictions. In the beginning two types of attachment get tested which are vertical to head and horizontal to head. Finding the most efficient attachment type whether fixed on a wall, a structure parallel to the head, or directly on the chair was the primary goal of this configuration. Instability and trouble maintaining the desired posture resulted from mistakes and performance problems in the flexion-extension range when utilizing the vertical position, which were discovered during the analysis. Consequently, the parallel-to-neck (horizontal) attachment position became the main focus of the design process. The horizontal arrangement is also simpler to incorporate with an office chair from a practical design standpoint. Consequently, the horizontal arrangement allows the head to stay in the intended target zone more successfully, to explain more when the spring put in vertical position the simulation run fails and neck reach impossible ranges. The results of the run

and the error of the vertical position can be seen in the figure 3.6 and 3.7.

Figure 3.6: *a) Horizontal position b) Vertical position*

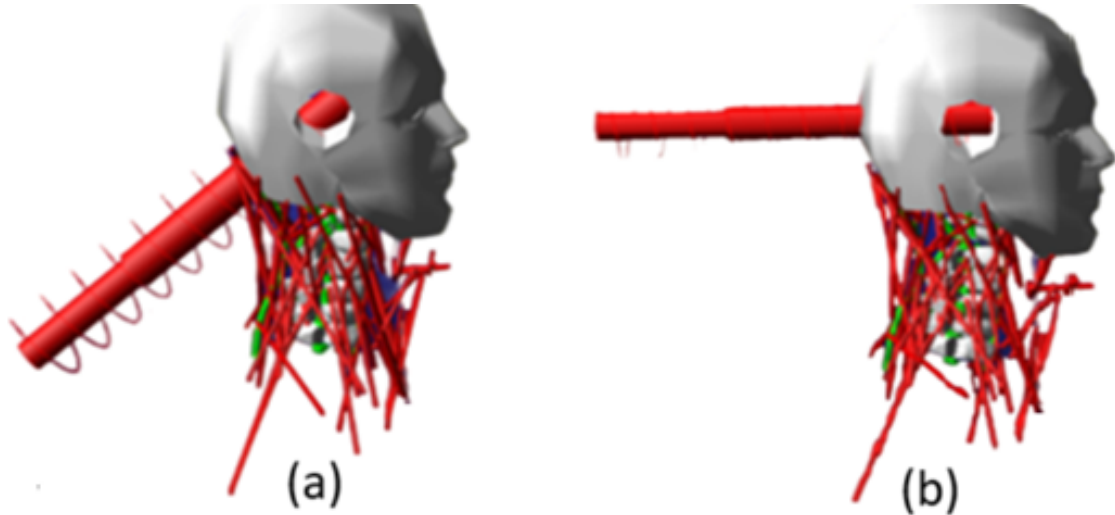
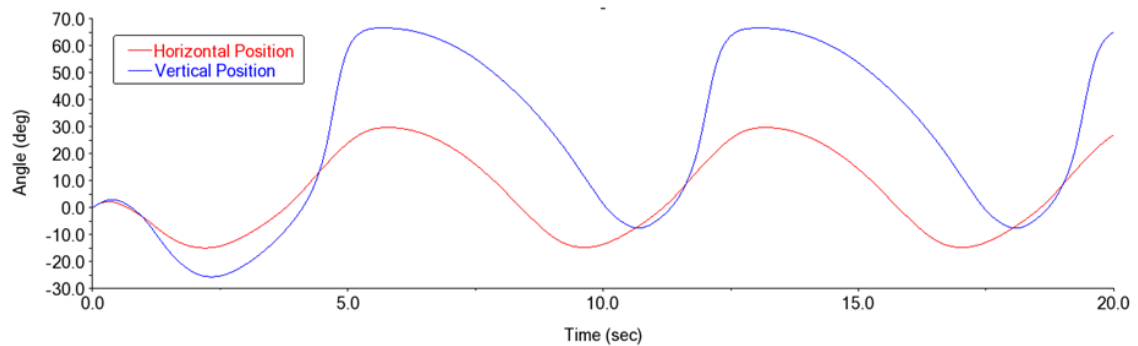


Figure 3.7: *RoM of horizontal and vertical positions*



Then, the top and rear of the head were considered as possible mounting locations in order to ascertain the best location for the springs in relation to head movement. The reasons behind this what kind of attachment will be more efficient. Rear head covers the head like pillow and the head covers like a hat or band. The spring placement as seen from the right side of the head is depicted in the figure. The best location for the springs to maintain the necessary range of motion (RoM) was determined to be at the top of the

head, based on the data displayed in Figure 3.8 and 3.9. From the simulation run top side of the head supports more efficiently than the back side of the head.

Figure 3.8: *a) Springs located at the back of the head b) Spring located on the top sides of the head*

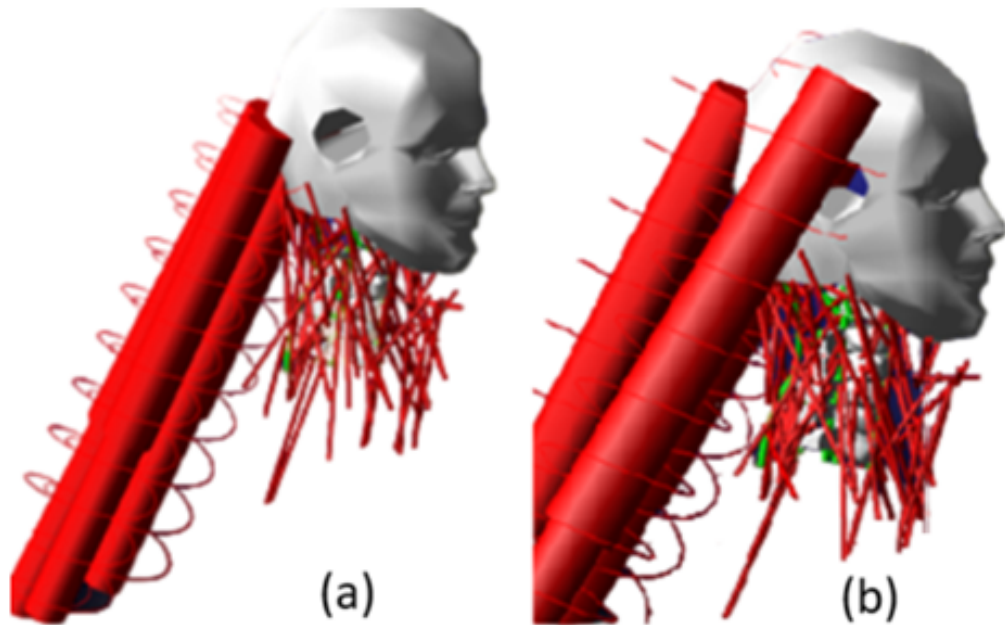
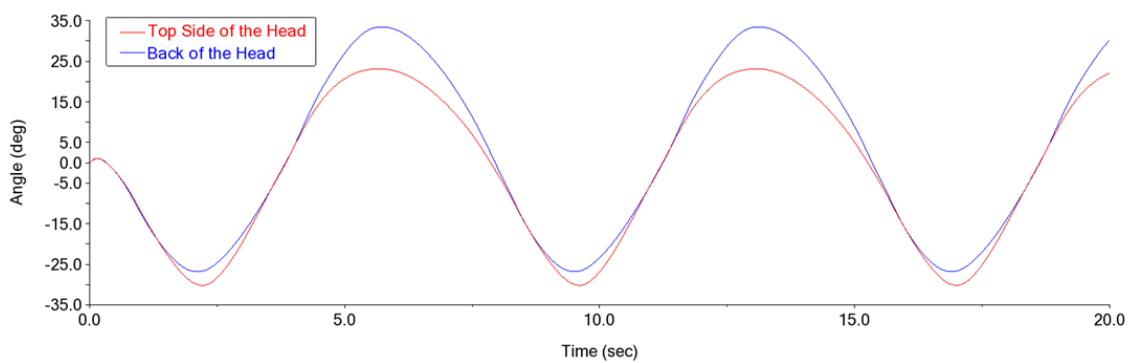


Figure 3.9: *RoM of top side of the head and back of the head*

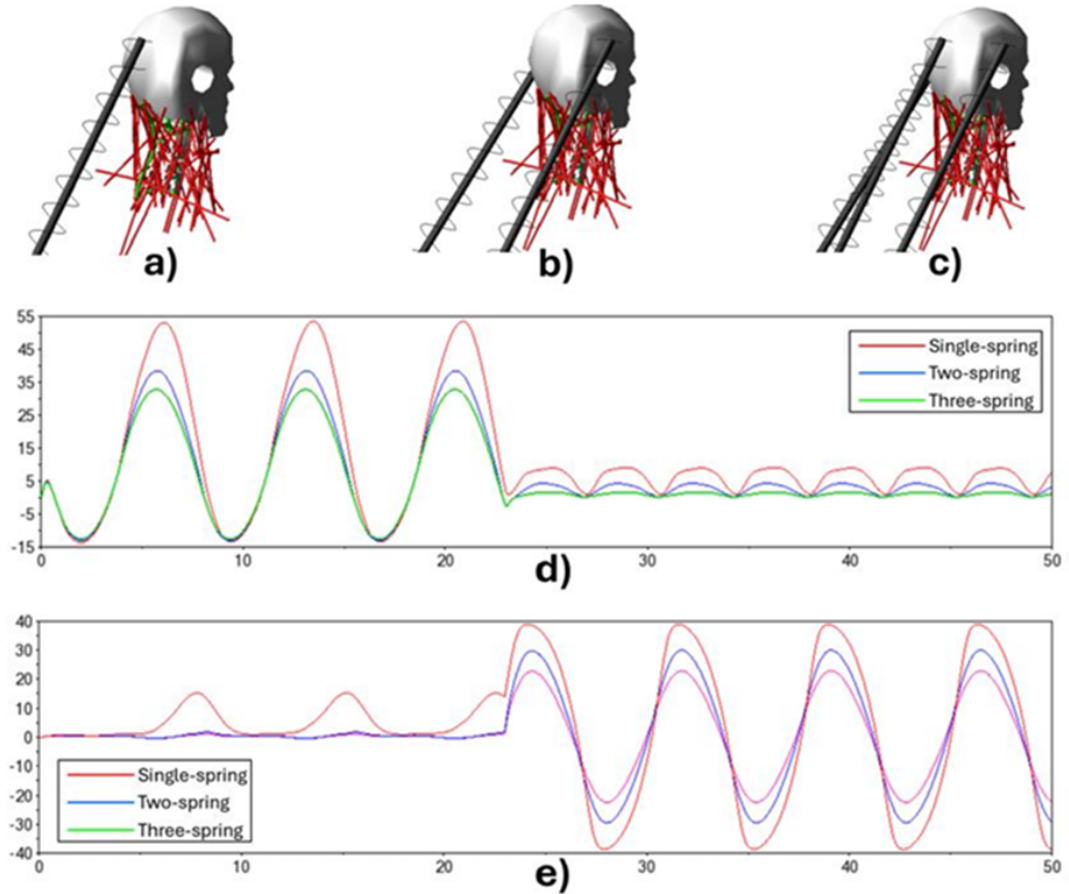


Then the amount of the springs get tested. The main motivation here is, to test one spring to see is there effect on lateral movement. Then the chair has two joint openings

which means if 2 spring is not enough the chair should be modified. Then, changing the amount of springs while maintaining a constant spring stiffness was the first step in the conceptual design process. As seen in Figure 3.10 three configurations a single-spring design, a two-spring design, and a three-spring design were assessed.

Because there was not enough room for a stable attachment to an office chair, configurations with more than three springs were not taken into consideration. Figures 3.10d and 3.10e show the flexion-extension and lateral flexion angle results that were obtained. The single-spring system was rejected since it was deemed inadequate for safe and regulated motion due to its exceeding of the intended range for both flexion-extension and lateral flexion. Despite allowing somewhat less lateral flexion, the two-spring structure was chosen as the final proposal because it produced a larger flexion-extension range of motion, which is more important for regular office work. The design decision placed a high priority on improving this range of motion because flexion-extension movements are more common in office duties.

Figure 3.10: Different configurations and their response a) single-spring configuration, b) two-spring configuration, c) three-spring configuration, and d) flexion-extension angle of three configurations e) lateral flexion angle of three configurations



As a result two a design with springs ergonomically fits to the concept design. Then single spring cannot be used. Since there is no effect on lateral movement. Therefore, three spring design offers the demands but as it mentioned before ergonomically it less efficient than the two springs. The work that made by the middle springs can be transmit to the side springs. That means 3 springs is not a necessity. Which is explained in the next part.

3.2.2 *Design of Mechanism*

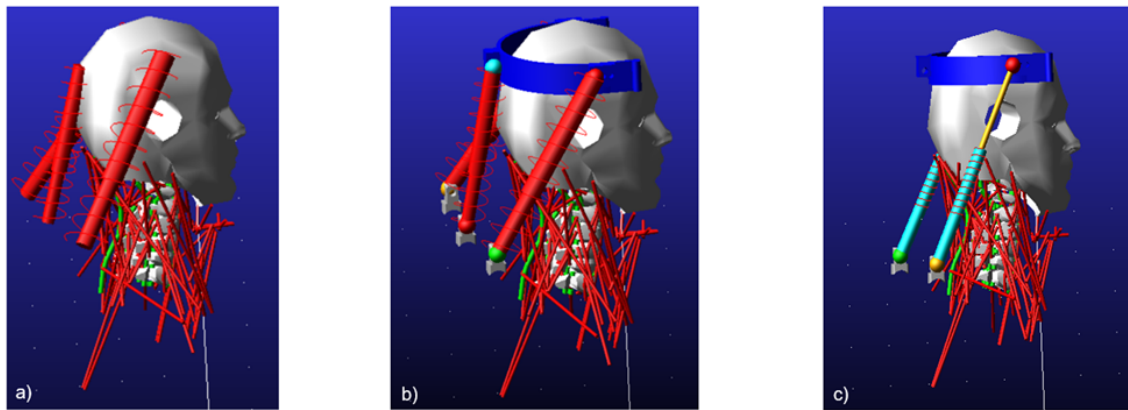
While defining the mechanism, three crucial factors were emphasized: the operation of the spring system, its application to the head, and its synchronization with head movements. The original idea for the spring mechanism included utilizing two compression springs on both sides, resulting in four springs overall. A piston-like design was suggested to efficiently compress these springs. To avoid buckling, the piston mechanism was designed to be enclosed within cylindrical tubes. Moreover, specially crafted caps were created for each end of the tubes to ensure adequate seating and secure placement of the springs inside the framework.

The objective of this spring mechanism was to attain maximum efficiency. In creating concepts for anchoring the system to the head, a headband-style design was deemed the most appropriate option. This headband is designed to attach the tube-mounted spring system to the head while offering a convenient framework for sensor installation, as it comfortably and securely encircles the head in a stable manner. To guarantee that the head and the mechanism function together seamlessly, the implementation of ball joints was recognized as crucial — particularly to permit unrestricted rotation. Ball joints provide a notable benefit for connections, as they can effortlessly connect to both the chair and the headband due to their standardized M-type inlets and outlets. Furthermore, a hook-like mechanism was suggested to fasten the system to the chair, utilizing the existing screw holes on the chair's structure for a stable and adaptable connection.

In the design process, the primary emphasis was on how the system could be mounted to both the chair and the head, with the connection intended via a ball joint link from point A to point B, illustrated in the figure below. Initially, a system with three springs was chosen, but as illustrated in the figure 3.11, when assembled, the third spring did not match the initial angle of the other two. This led to the central spring operating on its own, thereby heightening unpredictability in the system's performance. To tackle this,

the design was modified to a dual-spring setup, evenly reallocating the stiffness from the eliminated middle spring to the other two. Ultimately, the specific design features mentioned in the earlier paragraph were put into action, and the system was set up for DOE simulations to assess its performance and enhance the configuration.

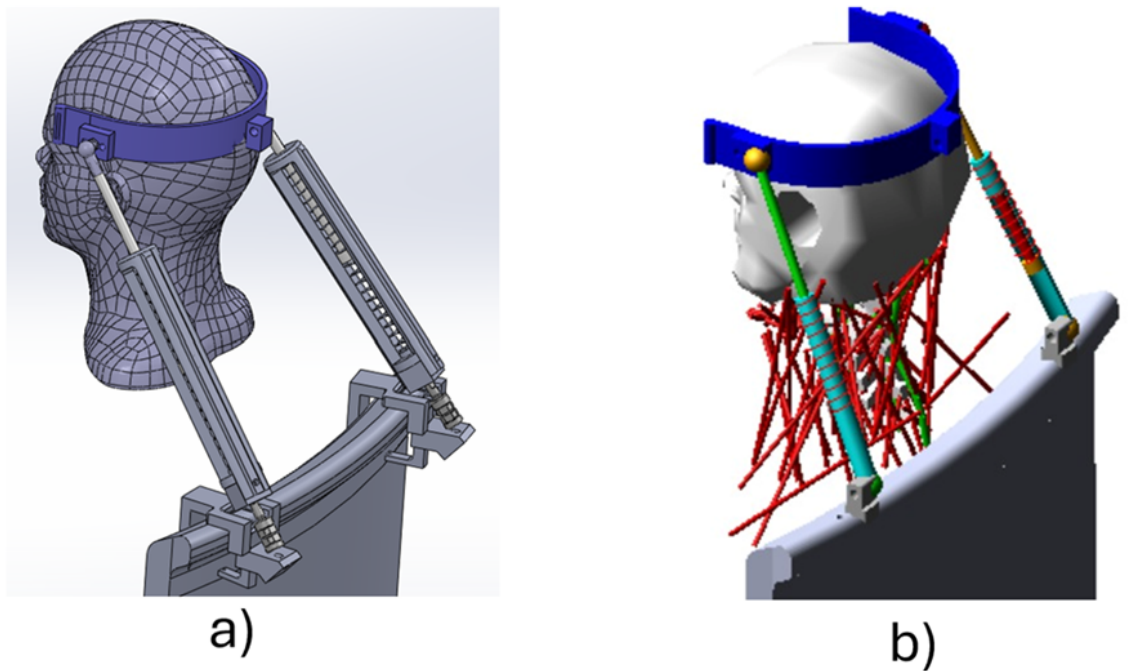
Figure 3.11: *a)First mechanism iteration, b)Second mechanism iteration c)Third mechanism iteration*



The conceptual design was then imported into SOLIDWORKS CAD software for further development, to carry out finite element analyses of the parts, and to create the suitable file formats for 3D printing. The changes implemented in this phase and the specific aspects of the components will be described thoroughly in the subsequent sections. The images below give a summary of the design in both its ADAMS simulation version and its CAD model format.

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Figure 3.12: a) *Design in CAD* b) *Design in ADAMS*



3.2.3 *Design Requirements*

The design planning stage of the prototype can start as soon as the neck model is ready to replicate all pertinent movements. Establishing precise design limitations is crucial to creating a successful prototype. Although a number of studies have been conducted on rehabilitation-oriented neck orthoses, as was indicated in the preceding sections, these devices usually limit or completely block neck movement instead of guiding it within safe, controlled ranges. This means that they do not offer particular biomechanical parameters appropriate for a preventive device. As a result, when determining suitable design constraints, the range of motion (RoM) of typical office workers becomes crucial. Angular RoM statistics for people with and without persistent neck discomfort have been reported by a number of studies.[61][80] These results were used in this study to establish target motion limits, which will avoid overuse and strain by safely limiting neck movements to

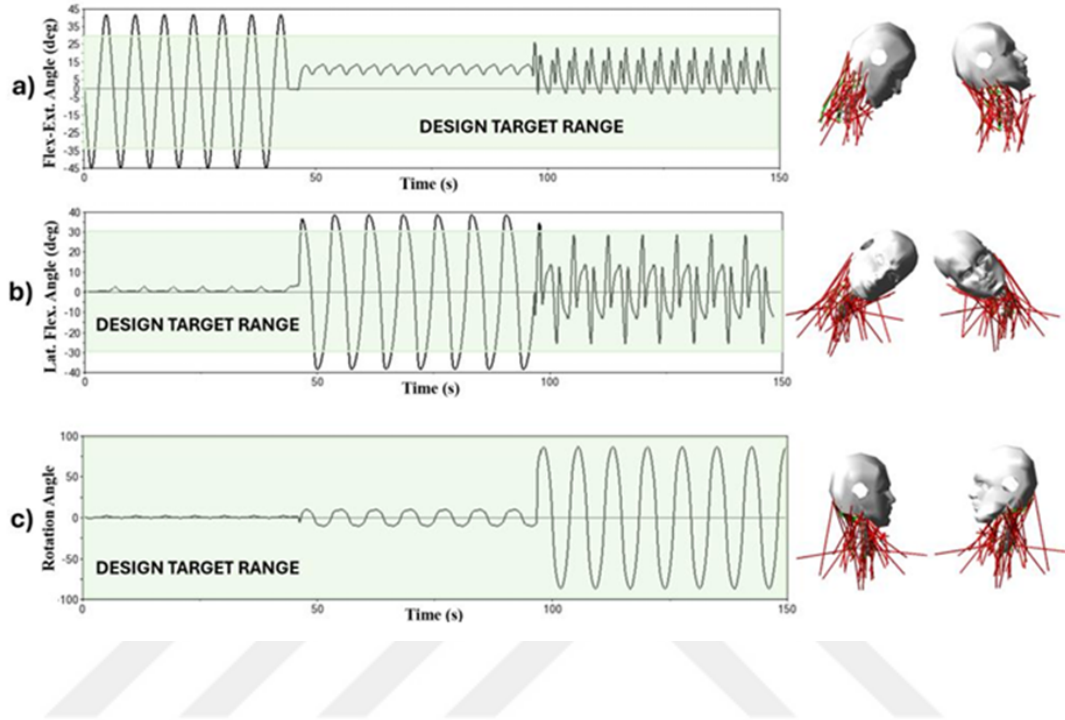
a range that persons with chronic pain typically cannot exceed. Although acceleration and neck velocity were also taken into consideration as possible limitations, they were eventually disregarded due to their negligible effect on the risk of harm for normal office workers[81]. As a result, these reference studies were used to determine the design restrictions for the device during neck movements, the design has been established in accordance with the typical neck movement limits, which are around 30 degrees of flexion, 35 degrees of extension, 30 degrees of lateral flexion, and 120 degrees of rotation compiled in the table 3.1 below. Then when the target ranges shown on the figure 3.13 it is clear to

Table 3.1: *Maximum neck movement angles by direction*

Direction	Maximum Angle (°)
Flexion	30°
Extension	35°
Lateral Flexion	30°
Rotation	120°

understand that rotation movements stay in the safe zone and this device should limit the flexion-extension and lateral flexion movement.

Figure 3.13: Target ranges of motion for the design objectives overlaid with a) neck flexion extension angle, b) lateral neck flexion angle, and c) rotation angle



As a result of this approach, the design limits in terms of RoM is set and after all these parameters plays a crucial role in the design and production of the setup.

3.3 Design of Experiment (DoE)

During the device's conceptual design stage, spring components were methodically examined and included into the neck support system using the ADAMS model. The goal was to ascertain the best location and spring stiffness levels to offer efficient support while preserving user comfort and guaranteeing the system's structural soundness. To achieve this, a number of simulations were run in the given design space to assess different spring designs and placements. In order to make sure that the observed effects were just the result of positional adjustments, the spring constant (K) was maintained constant while the quantity and placement of springs were changed. Using the ADAMS/Insight

program, a Design of Experiments (DOE) approach was used to optimize and analyze the mechanical system's behavior. This approach made it possible to precisely assess how various configurations affect the model's load distribution and motion patterns.

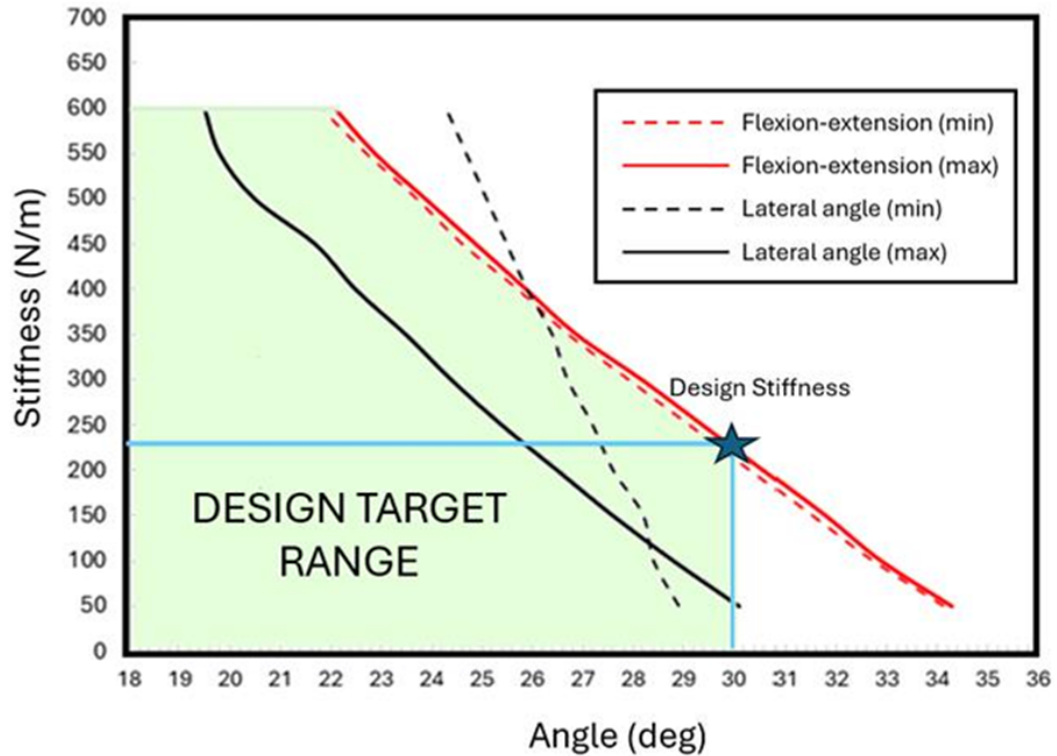
The necessary resistance levels for various neck regions were established as part of the investigation. In order to make sure that the springs adequately support normal head movements, the proper stiffness values for each region were determined by looking at biomechanical limitations and motion patterns. Multiple trials were conducted in an iterative manner; the design was progressively improved based on the simulation results, and finally the ideal configuration (Trial 5) was chosen to offer the optimum balance between flexibility and support.

Figure 3.14: DoE results

Trial	K stiffness	Flexion-extension angle (max)	Flexion-extension angle (min)	Lateral angle (max)	Lateral angle (min)
Trial 1	50	34.3146	34.1042	30.128	28.9016
Trial 2	100	32.9717	32.7317	28.8431	28.4563
Trial 3	150	31.7552	31.5025	27.6286	28.2162
Trial 4	200	30.559	30.2905	26.48	27.5819
Trial 5	250	29.4001	29.1052	25.3999	27.1528
Trial 6	300	28.2234	27.8969	24.3867	26.7289
Trial 7	350	26.9508	26.7102	23.5482	26.3982
Trial 8	400	25.9208	25.6823	22.5482	25.8962
Trial 9	450	24.8662	24.6092	21.7138	25.4928
Trial 10	500	23.9625	23.7265	20.4901	25.0982
Trial 11	550	22.9635	22.7225	19.8011	24.7042
Trial 12	600	22.1425	21.7795	19.5071	24.3205

Then, to increase the understanding the DoE results in graphics in the figure 3.15 below. The green area shows the regions that offer the design target range. Furthermore the point mentioned as a star is the optimum result comes from the DoE.

Figure 3.15: *DoE results in graphics*



3.4 Setup Design

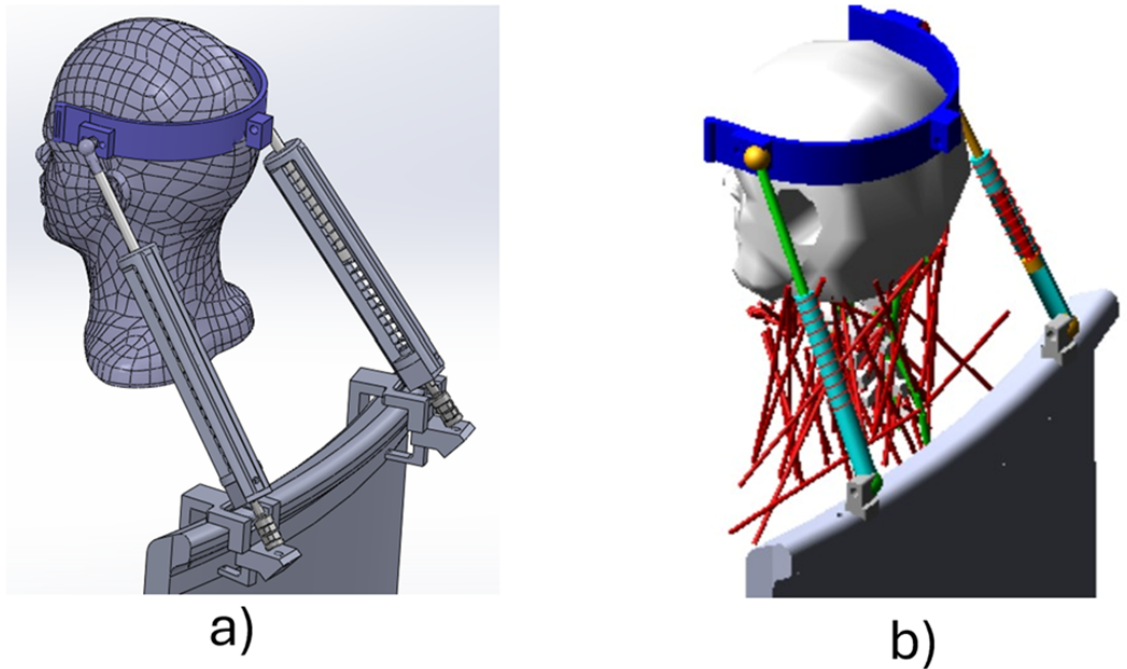
In this designing process, all other components were specifically built and modeled for this system, with the exception of some externally supplied pieces. In a nutshell, the general design and modeling methodology is in order to assess the system's mechanical behavior under realistic circumstances, the design was initially modeled and simulated in ADAMS as it mentioned in the previous parts. Ball joints, which enable movement across six DoF and provide a greater range of motion, were used to connect the essential parts. Springs placed between the top and bottom ball joints provide stability for this improved motion. The springs work like pistons to provide controlled and smooth movement; they are housed inside cylindrical tubes and fastened to rods.

Following the preliminary simulations, the design was imported into a CAD environment, where the component interfaces, ball joints, and connections were accurately defined to meet real-world production needs. The final step involved integrating the entire apparatus with a seat assembly, preparing it for practical use. From virtual simulation to CAD detailing and assembly, this methodical process guarantees that the item will operate at peak efficiency and versatility when put into real-world use. Accurately describing each component's dimensions, placement, and connection types throughout the CAD design process helped to match real-life situations.

Following the implementation of the finalized CAD model in ADAMS, the ideal spring constant was identified in order to verify that the system functions as intended, as previously mentioned.

The chair and ball joints in this drawing were not specially made; rather, the technical drawings of the ball joints were taken straight from the websites of the manufacturers and added to the assembly with due credit. It's not being used without permission. The chair's dimensions were measured and precisely entered into the software to create the model. The system's head model was taken from a third-party website and incorporated into the project with the original creator's consent[82].

Figure 3.16: a)CAD Model b)ADAMS Model



The CAD model was built using SolidWorks and is illustrated in the figure below as an exploded view with numbered components. To explain it briefly, the main parts of the system, each intended to perform a distinct role within the larger mechanism, are depicted in the figure. The headband, which firmly fastens the system to the user's head, is the first part. The 54-gram headband, which is made of PLA material, also material selection will be discussed in the next parts. is an essential connecting point.

The ball joint, DIN 71802 that connects the headband to the mechanical assembly is the second part. These joints, which weigh 22 grams each, serve as a bridge connecting the head to the support system, enabling mobility.

The metal rods, which are composed of stainless steel and have dimensions of 10 mm in diameter, 200 mm in length, and 105 grams in weight, make up the third component. These rods' lower ends have a cylindrical, piston-like tip that is 10 mm long and 16.5 mm in diameter. It has a flat surface and a safe groove or recess to firmly anchor the springs,

while the upper ends attach to the ball joint. The weight of each rod is 55 grams.

The springs, the system's fourth component, give it the force and balance it needs. A total of four springs, each weighing five grams, have a length of 200 mm, a spring constant of 250 N/m, and other characteristics that were established through simulation optimization. The fifth component consists of PLA tubes that contain the springs. With an outside diameter of 20 mm and an interior diameter of 16.5 mm, each tube has a length of 250 mm. In order to guide the springs and stop off-axis movement when in use, the tubes are closed at the top and open at the bottom.

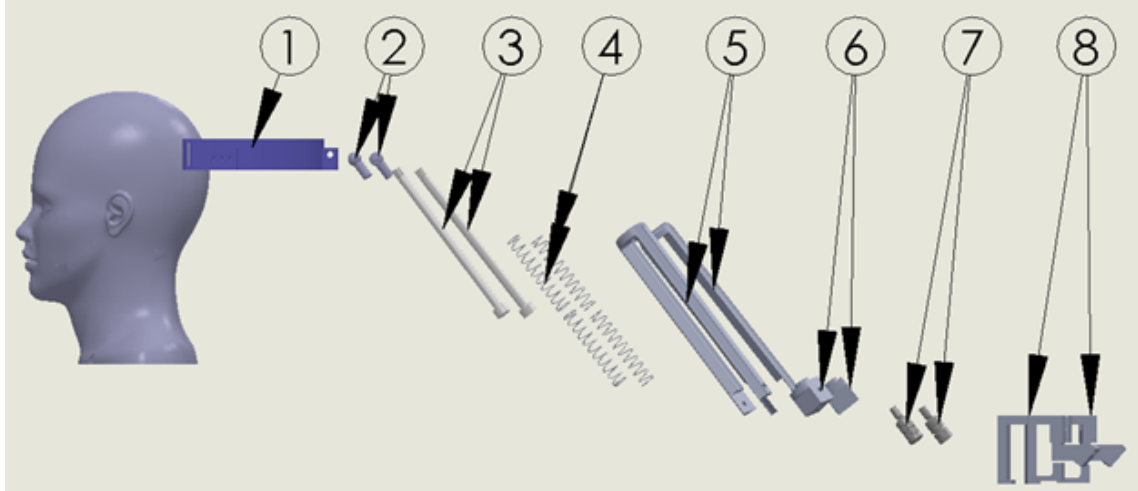
The sixth component, tube caps, are used to secure the springs inside the tubes. Each 11-gram cap, which is also composed of PLA, serves as a closure to join the tube to the subsequent component in the system.

A second ball joint, GN-782M6, which has a greater range of motion than the first, makes up the seventh part. This 15-gram joint guarantees that the system can operate within the necessary degrees of freedom.

The tailored connection, which firmly fastens the system to an office chair, is the final part, number eight. This connector, which weighs 38 grams and is made of PLA, guarantees a strong connection between the chair and the mechanical system.

About 546 grams is the total weight of the entire system. When combined, these thoughtfully crafted parts form a useful, adjustable neck support system that puts user comfort, seamless motion, and structural integrity first. Exploded view of the design can be seen in the Figure 3.17

Figure 3.17: Exploded view of the design: head band (1), ball joints (2), rods (3), springs (4), tubes (5), tube caps (6), ball joints (7), connectors to the chair (8)



3.4.1 Material Selection

Before examine the parts detailed. The metarial selection of the materials are crucial. PLA was chosen as the preferred material because the device is intended to be produced via 3D printing. PLA is a biodegradable thermoplastic made from sugarcane or corn starch, two sustainable plant-based materials. Although it provides a notable benefit in terms of environmental sustainability, its efficient decomposition necessitates industrial composting facilities[83]. With a nozzle temperature of 180–220 °C and usually no heated bed or just a modest bed temperature (0–60 °C), it is renowned for being easy to print. It is especially useful for creating prototypes because of its low propensity to warp, which makes it easy for novices to use[84]. High stiffness and brittleness are characteristics of its mechanical qualities. Its impact resistance is limited, with an elongation at break of about 4%, although having a tensile strength (60 MPa) that is equivalent to Acrylonitrile Butadiene Styrene (ABS). Because of this, PLA performs poorly in sections that need to be flexible[85]. Heat resistance is a known drawback; its usage in automotive or other high-temperature applications is limited because softening usually starts beyond

60°C[86]. From that in the part number 3 was made by stainless steel to avoid the risk of the lateral force and the heat produce due to the friction. Because of its biocompatibility, it can be utilized in items that come into contact with food and in medical implants, particularly in certified versions that adhere to ISO 10993 criteria[87]. PLA's broad use in prototyping and instructional projects is further supported by the fact that it is less expensive than ABS[88].

3.4.2 FEA of critical parts

The structural integrity of the components was then assessed using a deformation analysis when a load of 50 N—roughly the weight of a human head—was applied. The various components' amounts of deformation under this loading scenario are depicted in Figures 3.18 and 3.19 . The findings show that the deformation values stay within allowable bounds, with the majority of components showing all values below 0.01 mm. This satisfies the design specifications and validates that the system can securely support the anticipated load.

In particular, both the chair attachment and the headband display deformation levels ; 0.03 mm for the deformation depicted in Figure . Localized stress is indicated by the deformation, which is mostly focused at the joint attachment points of both components. The rod exhibits a 0.06 mm deformation under static loading conditions in Figure . The component's structural stability under the applied load is confirmed by the spring placer's minimal deformation, which is measured at 0.002 mm.

The plastic deformation limit of PLA, which normally varies between 0.5% and 5% strain based on the particular material qualities, is significantly lower than these deformation limits. The findings guarantee that the material maintains its structural integrity under the imposed load without suffering any irreversible deformation by confirming that the measured deformations stay within the elastic range. PLA's suitability for usage in this design

application is further validated by its alignment with its known mechanical behavior.

To continue with the stress values, the values reach maximum 3 MPa for head band but more importantly, the critical lines can be seen clearly. This is important since not just applying on the experiments, during the montage process these critical regions don't push too much to cause overloading and failure on the parts.

As a result, the von Mises stress values found in the investigation are significantly lower than the normal yield strength of polylactic acid (PLA), which, depending on the material grade and print settings, typically ranges from 50 MPa to 70 MPa. This demonstrates that, under the imposed load, the components function safely within the elastic stress limits of the material. Because of this, the design eliminates the possibility of yielding or irreversible deformation, guaranteeing that the PLA components will continue to function and have structural integrity. This result further supports the selection of PLA for this application by showing that the stress distribution is in line with the material's mechanical capabilities.

Figure 3.18: Stress Analysis of a) rods b) head band c) tube caps d) connectors to the chair.

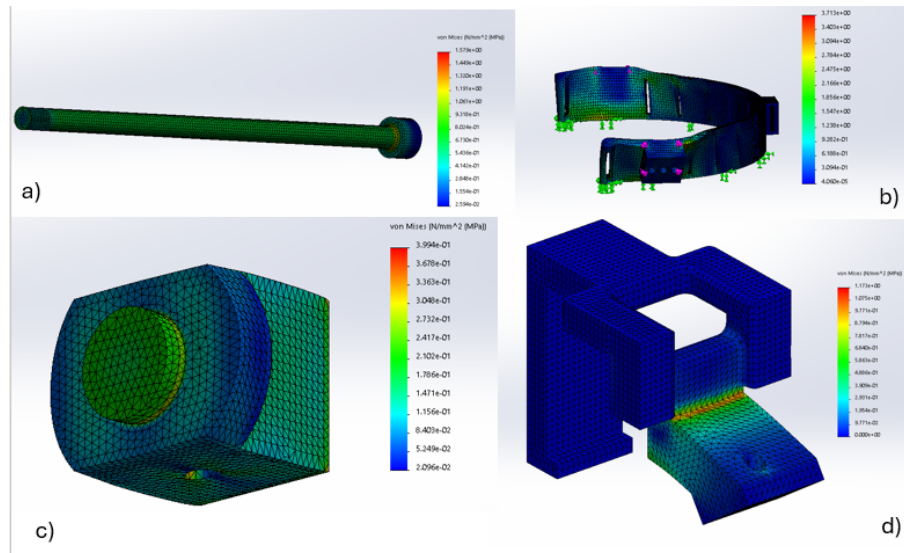
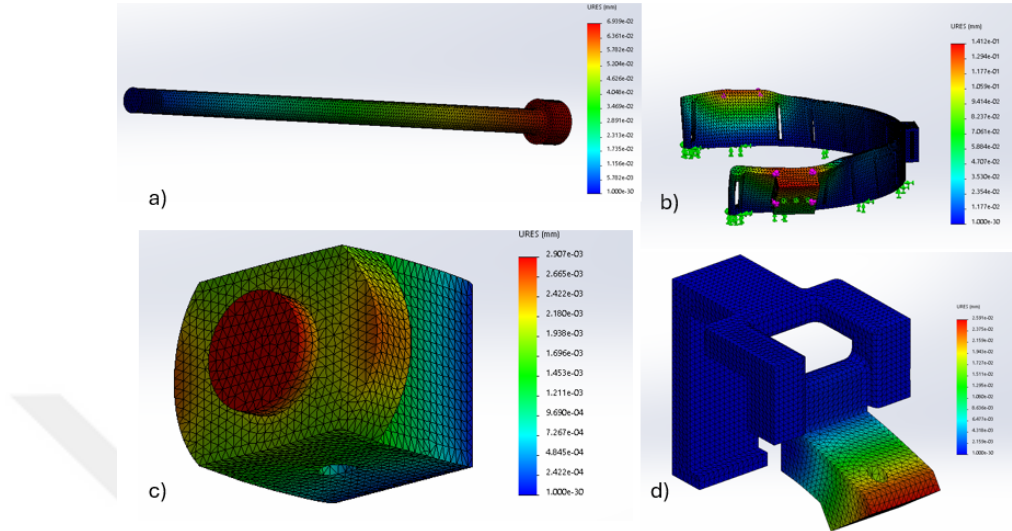


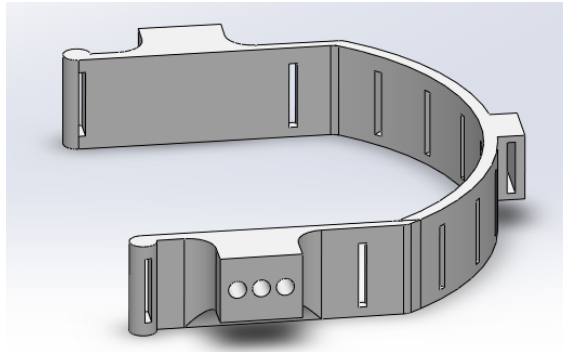
Figure 3.19: *Displacement Analysis of a)rods b)head band c)tube caps d)connectors to the chair.*



3.4.3 Detailed Explanation of the Parts

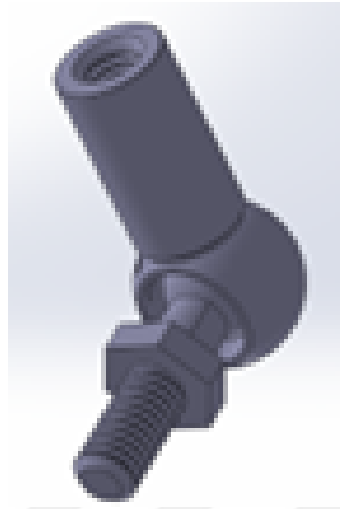
In this part the parts of the NACK setup explained orderly. 1- Head Band: The headband acts as the main connection between the user and the supportive neck assistance system. It is crafted to securely and comfortably fasten to the user's head, evenly distributing forces from the support system to prevent localized pressure areas. The headband is made from PLA through 3D printing, making it lightweight, simple to customize, and sturdy enough for frequent use. It has an approximate weight of 54 grams, adding little extra weight to the user's head. The headband has uniquely crafted mounting points that attach to the upper ball joints of the system, which in turn connect to the spring-and-rod mechanism that offers dynamic neck support. This part is essential in transferring support forces from the spring mechanism to the head.

Figure 3.20: *Head Band*



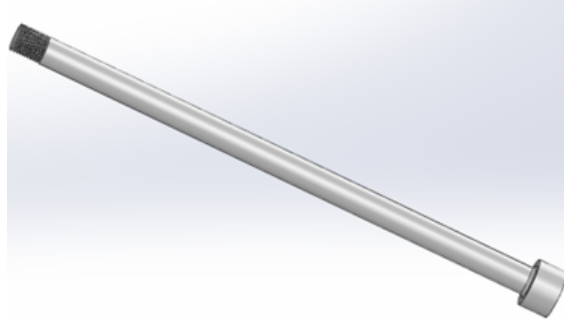
2- DIN 71802 (Ball Joint): The ball joint specified by DIN 71802, utilized in the NACK project, is a standardized joint with adjustable angles intended for fluid multi-directional movement in mechanical connections. Usually made from carbon or stainless steel, it includes a hardened precision steel ball contained within a metal casing that enables smooth, low-friction motion. This design allows for an angular deflection of around 15° to 20° for each axis. A locking nut or comparable fastening component holds the joint securely during assembly. DIN 71802 ball joints are commonly employed for control rods, linkages, and adjustable mechanical connections where flexibility and safe load transfer are essential. In the NACK system, this element is essential for facilitating natural neck motion while consistently transferring forces from the spring mechanism to the headband.

Figure 3.21: *DIN 71802*



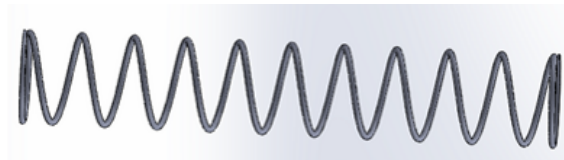
3- Metal rod: In the NACK project, the metal rods serve as essential structural components that link the headband to the spring mechanism and convey the support force to the user's neck. These rods are made from stainless steel, selected for its exceptional strength, resistance to corrosion, and durability when subjected to repeated loading conditions. Every rod features a diameter of 10 mm and a length of 200 mm, ensuring a balance between strength and manageable weight. The top end of each rod is linked to the headband through a ball joint, permitting unrestricted rotational motion aligned with the system's multi-directional support structure. At the bottom of each rod, there is a uniquely crafted piston-like tip with a length of 10 mm and a diameter of 16.5 mm that acts as a reliable point for fastening the spring. This suggestion features a flat area and a cut groove or indentation to firmly secure the spring, guaranteeing regulated flexing during head and neck motions. Each rod has a weight of about 105 grams, enhancing the system's overall stability while not introducing excessive bulk. The metal rods collectively form the crucial structural connection between the user's head and the chair-mounted support system, ensuring that the device offers reliable preventive support while permitting natural neck movement.

Figure 3.22: *Metal Rod*



4- Springs: In the NACK project, the springs serve as the essential functional elements that provide the adjustable support force necessary for sustaining optimal neck posture. Totally 4 compression springs are positioned between the headband and the lower attachment points to evenly distribute the weight of the head and lessen muscle strain during extended sitting. Every spring possesses a spring constant of 250 N/m, a figure established through simulation research to offer adequate resistance while permitting natural head movements. The springs measure 200 mm in length and each weighs approximately 5 grams, guaranteeing that they contribute negligible weight to the system while offering efficient force storage and retrieval.

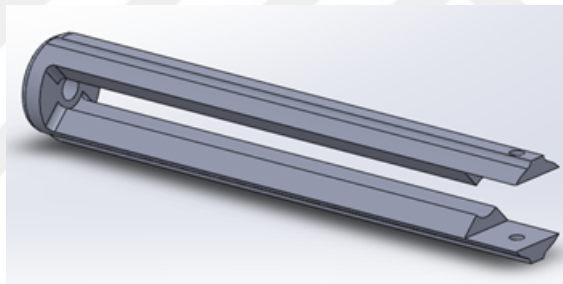
Figure 3.23: *Compression Spring*



5- Spring Tubes: In the NACK project, the spring tubes act as protective and guiding casings for the compression springs, making sure they function smoothly and stay correctly aligned during operation. Every tube is produced from PLA through 3D

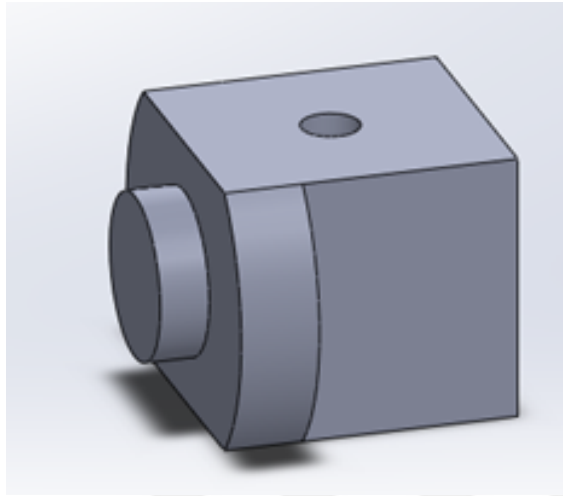
printing, enabling a lightweight and sturdy structure customized to the device's design limitations. The tubes possess an outer diameter of 20 mm, an inner diameter of 16.5 mm, and a length of 250 mm, offering ample space to fit the complete deflection range of the springs without leading to buckling or side movement. The upper end of each tube is sealed to secure the spring's top, while the base stays open for connection to the piston-like tip of the metal rods. Custom tube caps are utilized to secure the springs and prevent them from slipping or becoming misaligned, effectively anchoring the springs while connecting the tubes to the rods and ball joints underneath. By ensuring exact alignment, the spring tubes are essential for transmitting the spring force effectively to provide support.

Figure 3.24: *Spring Tube*



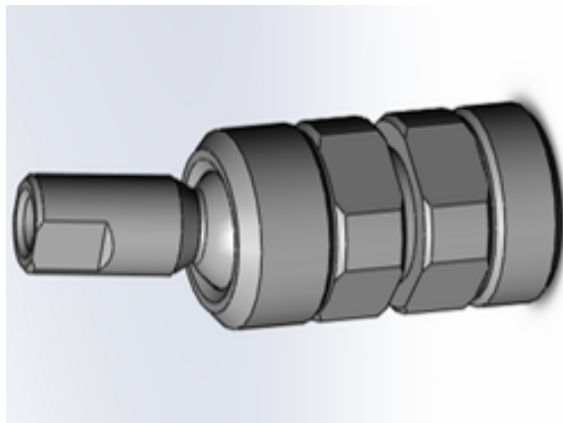
6- Tube Caps: In the NACK project, the tube caps are tiny yet crucial parts that hold the springs inside the spring tubes and link them securely to the overall support structure. Every tube cap is 3D fabricated using PLA, guaranteeing it is lightweight while also being sturdy enough to uphold the structural integrity of the spring housing. The main role of the caps is to seal the top of the spring tubes, stopping the springs from moving out of place during bending.

Figure 3.25: *Tube Cap*



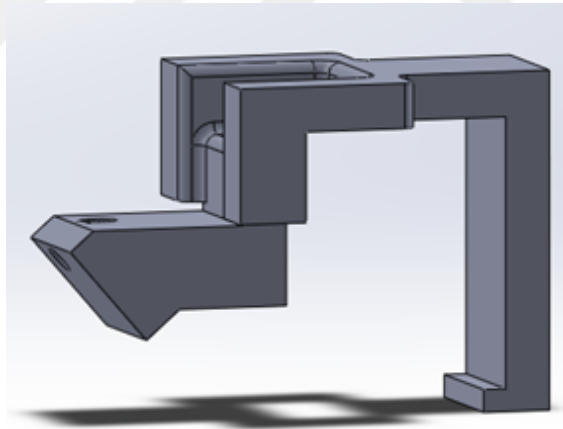
7- GN782 (Ball Joint): The GN 782 Ball Joint (produced to Elessa+Ganter specifications) is a joint that allows for angle adjustment, intended for accurate alignment in mechanical systems. It includes a sturdy casing made of zinc-coated steel or corrosion-resistant stainless steel (AISI 304), combined with a hardened steel or stainless-steel ball that rotates inside a precision spherical bearing socket. This design allows for seamless angular adjustments of 15° – 18° in every direction.

Figure 3.26: *GN782*



8- In the NACK project, the tailored connectors — known as the bottom connectors — are crucial for securely fastening the complete assistive system to an office chair. These connectors, produced through 3D printing with PLA, are crafted to align with the contours and attachment points of a standard office chair's backrest or frame, guaranteeing a stable and discreet fit in actual workplace settings. The tailored connectors serve as a dependable anchoring point for the lower ends of the spring tubes and rods, ensuring the entire mechanism remains correctly aligned during operation. This strong connection guarantees that the spring-based system stays in place while the headband and upper framework can move with the user's natural neck motions. These connectors, weighing around 38 grams each, are lightweight but structurally adequate, preserving the device's ergonomic function without increasing bulk or sacrificing the chair's inherent comfort.

Figure 3.27: *Connector part*



3.5 Manufacturing and Montage Process

In this section the manufacturing and montage process of the prototype examined. During the 3d manufacturing process, the parts extracted to .stl file type and be prepared to the 3d printing process. The Cura software used during 3d printing. The Ultimaker 2 Extended 3D printer was chosen due to its accuracy and dependability in creating useful

polymer parts. A 0.4 mm brass nozzle was utilized during the printing process, and the ideal temperatures for Ultimaker's regular PLA filament were set at 210 °C for the nozzle and 60 °C for the build plate. Ultimaker PLA was selected as the material because of its reliable quality and easy printability; it meets the mechanical specifications of the intended neck support system with an elongation at break of 6%, an E-modulus of roughly 3.5 GPa, and an average tensile strength of about 61 MPa. These specifications made sure that the printed components had the necessary structural stability and dimensional accuracy while still being simple to assemble and post-process.

Mounting the components in order, beginning with the bottom section and working up to the headband, was the first step in the construction process. Insert nuts, which offered sturdy, dependable fastening points and guaranteed that the parts could be repeatedly assembled and disassembled without causing damage to the printed material, were used to firmly join the 3D-printed components. Throughout the whole setup, exact alignment and structural stability were ensured by this modular method. To guarantee consistency and repeatability in the testing procedure, the same assembled configuration was used for all ensuing studies. At the end in the figure the complete setup can be seen.

Figure 3.28: *a)First Prototype b)Prototype with modifications*



4. DESIGN, DEVELOPMENT AND FUNCTIONAL EVALUATION OF NECK ASSISTIVE CERVICAL KIT (NACK)

In this part the making of the experimental procedure will be explain. Then the experimental process demonstrates with support of the figures from the experiments. Moreover, the results of the experiment examined. Lastly, the results and the hypothesis justified.

4.1 Experiment Tools

In the experiments EMG sensors used for the muscle activization, IMU sensor used for the getting real time data such as acceleration, velocity, position and Euler angle. Then, the Bournemouth questionnaire asked to participants to evaluate the starting state of a number of variables, including functional capacity, emotional state, and pain level. Lastly, pain intensity of participant was measured with the Numeric Pain Rating Scale (NPRS 101, range: 0–100)

4.1.1 *EMG measurements*

Electromyography (EMG) sensors are devices that monitor the electrical activity generated by skeletal muscles during contraction. When a muscle contracts, electrical signals resulting from muscle fiber depolarization can be monitored at the skin's surface or directly within the muscle using electrodes. Surface EMG (sEMG) sensors, such as those used in ergonomics and biomechanics research, are non-invasive and involve applying adhesive electrodes to the skin above the muscle of interest. These sensors detect small voltage changes (usually in the microvolt to millivolt range) and transform them into signals that can be recorded, analyzed, and used to determine muscle activation, exhaustion,

coordination, or movement patterns. So, for this investigation, emg sensors were perfect for conducting trials.

EMG measurements were performed after reviewing relevant literature on neck muscles. Based on an assessment of the most optimal measurement sites, it was concluded that, for example in one research, splenius capitis (SC) muscles examined as a superficial muscles[89]the upper trapezius (UT) and sternocleidomastoid (SCM) muscle groups are the best fit for this purpose.

The reasons for choosing these two muscle groups are also shown in the figures below. The chosen neck muscles, the SCM in the front and the UT in the back, comprise the majority of the major superficial muscles in these areas. Because they are among the outermost muscle groups, they provide major measurement benefits.

The upper trapezius (UT) muscle is a critical superficial muscle in the posterior neck and upper back that controls scapular and cervical mobility. The most visible part of the trapezius, it starts from the occipital bone and the nuchal ligament and enters into the lateral third of the clavicle and the acromion of the scapula. Its principal tasks include raising the scapula (as seen when shrugging the shoulders), assisting in neck extension, and contributing to lateral neck flexion[90].

The sternocleidomastoid muscle (SCM) is a conspicuous superficial muscle in the anterior neck that stands out when the head is moved or inclined. It has two heads: the sternal head attaches to the manubrium of the sternum, and the clavicular head emerges from the medial part of the clavicle. The mastoid process of the temporal bone and the superior nuchal line of the occipital bone are where both heads merge and implant. Its main actions are ipsilateral lateral flexion (tilting the head to the same side) and contralateral rotation (turning the head to the opposing side). When both SCM muscles contract simultaneously, they contribute in neck flexion and forceful inspiration by lifting the sternum.[91]

Furthermore, a review of the relevant literature assisted in determining which exact parts of these muscles the EMG electrodes should be placed on for the best signal quality. This placement adheres to Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles (SENIAM) guidelines, which are the standard protocol for upper trapezius EMG measurements. The electrode placement was standardized by identifying the midpoint between the spinous process of the seventh cervical vertebra (C7) and the acromion (the shoulder's lateral tip). This location allowed for the most accurate detection of muscle activation along the UT's major fiber direction. The inter-electrode distance was fixed at 2 cm, and the electrodes were oriented parallel to the muscle fibers to improve signal accuracy and reduce crosstalk from adjacent muscles.[92] Surface EMG electrodes were implanted along the muscle belly of SCM muscle, parallel to the muscle fibers, at a point about one-third of the distance between the mastoid process (behind the ear) and the sternal notch. To ensure clear signal acquisition and minimize cross-talk from surrounding muscles, electrodes should be placed directly over the muscle belly, parallel to the fiber direction, and 2 cm apart.[93][94]The figure below demonstrates the application on sternocleidomastoid muscle (SCM) and UT.

Moreover, The NACK project defines Maximum Voluntary Contraction (MVC) as the maximum amount of force or muscle activity that a person may voluntarily produce during a certain isometric contraction. It is used as a common reference point in EMG research to compare muscle activation levels across different situations. MVC measures were taken for each participant and for each target muscle, namely the SCM and UT muscles on both sides. Participants were instructed to contract each muscle as vigorously as possible while maintaining a controlled posture in order to obtain consistent and accurate data. During these Maximum Voluntary Contraction (MVC) trials, the peak EMG amplitude indicated that muscle was fully activated. This standardization permitted a clear and comparable investigation of how the assistive device affected neck muscle load during

daily activities. As a result, in order to test the MVC for SCM, participants rotated their heads as much as possible. The UT muscle was next worked on by abducting both upper arms against opposition[63]

The ensuing tests used BiosignalsPlux EMG sensors to record electromyographic activity. To ensure data acquisition reliability and consistency, sensor insertion was done in accordance with published literature protocols. Standard ECG (electrocardiogram) electrodes were modified (reduced to size) and adapted for use in an EMG setup for neck electrode insertion. The key motivation for choosing electrocardiogram (ECG) electrodes was entirely economic, since they provided a cost-effective alternative without considerably compromising signal quality for the study's objectives. This method exhibits a realistic adaption of existing materials to satisfy experimental requirements while retaining scientific rigor. This study used a PLUX biosignalsplux EMG sensor to evaluate the electrical activity of the neck muscles. This surface electromyography (sEMG) sensor detects skeletal muscle impulses using three electrodes—two input electrodes and one reference electrode—adhered to the skin with adhesive pads. The sensor detects muscular activation signals within a ± 1.5 mV range and frequency bandwidth of 20 Hz to 450 Hz, which covers most voluntary muscle activity. To accurately record signals, a sampling rate of at least 1000Hz is recommended. The system generates low noise levels of 1-2 μ V root mean square (RMS) Root Mean Square (RMS), resulting in excellent signal quality.

The compact, wireless module easily connects to PLUX's biosignalsplux platform, which supports real-time data transfer and multi-channel recordings, making it perfect for rehabilitation, muscle function testing, and biomechanical studies. The sensor is also completely compatible with open software development kits (SDKs) for MATLAB®, LabVIEW®, and other research tools, allowing for flexible data processing and integration into customized experimental workflows.

Therefore, for the filtering process, First, the raw EMG data recorded during tasks was band-pass filtered between 20 and 490 Hz using a fourth-order Butterworth filter. This frequency range was chosen to avoid motion artifacts and high-frequency noise while maintaining the crucial EMG signal bandwidth. After band-pass filtering, the signals were full-wave rectified to determine the absolute value of the EMG amplitude. To recover the muscle activation envelope, the rectified signals were sent through a second low-pass Butterworth filter with a cutoff frequency of 6 Hz. This smoothing process produced the linear envelope that depicted the overall muscle activity pattern throughout time. The filtered and enveloped EMG data were then standardized to each muscle's maximal voluntary contraction (MVC) values, enabling for comparisons of muscle activation levels with and without the NACK device.

4.1.1.1 IMU Sensors

The inertial measurement unit (IMU) sensor was selected using a highly rigorous methodology. A examination of the literature indicated that a variety of approaches have been used to assess neck mobility. Firstly, In this study, neck posture was assessed using the Craniovertebral Angle (CVA) approach. To document this, two physical markers were placed on each participant: one at the spinous process of the C7 vertebra and the other at the tragus of the ear. A webcam was positioned laterally to capture continuous video footage of the participants' posture as they worked on computer-based document activities. The recorded photos were processed in real time using a Python program based on OpenCV, which automatically recognized the positions of the markers and calculated the CVA constantly throughout the 50-minute session[68] In the investigation, a soft cap was used to put a 3-axis gyroscope sensor (MPU-6050) on the participant's head. This sensor assessed real-time head flexion angles (sagittal plane tilt) by integrating gyroscope

and accelerometer signals to calculate head position relative to the vertical. These experiments demonstrate how a wearable sensor can be employed, and the sensors used in the final step were IMU sensors.[67] Lastly, in a different study three compact IMU sensors were attached to the left, right, and back of the neck, surrounding the cervical region, allowing for comprehensive tracking of head movements across the frontal, sagittal, and transverse planes. These sensors collected orientation data in real time, providing estimations of head tilt (pitch), lateral bending (roll), and rotation (yaw). To ensure precise tracking, raw accelerometer and gyroscope inputs were combined with a Kalman filter, resulting in smoothed and trustworthy estimates of each IMU's roll, pitch, and yaw angles. The entire system ran on a real-time microcontroller, which continuously sampled IMU data and classified the user's posture into defined risk levels. Then it warns the user to stay the user in a correct posture.[95] As a result we decided to use IMU sensor to precisely quantify neck movements, an IMU sensor is used, which commonly consists of a triaxial accelerometer, gyroscope, and, in certain cases, a magnetometer. The IMU was firmly fixed to the participant's head or a head-mounted device, allowing it to collect three-dimensional kinematic data in real time. The accelerometer measured linear acceleration during dynamic motions (for example, quick head turns), while the gyroscope tracked angular velocity, allowing sensor fusion algorithms (such as complementary filters or Kalman filtering) to calculate neck flexion, extension, lateral bending, and rotation angles. By combining these signals, the IMU provided objective metrics such as range of motion (ROM), movement smoothness, and velocity, which were then synchronized with EMG data to correlate muscle activation patterns with specific neck movements.

Therefore the IMU sensor checked in the industry and LPMS B2 IMU sensor selected. The LPMS-B2 by LP Research is a t wireless 9-axis IMU with AHRS designed for precise

motion and orientation monitoring. The device includes a 3-axis gyroscope, accelerometer, and magnetometer, as well as temperature and barometric pressure sensors, for accurate altitude estimation. The sensor uses a 32-bit digital signal processor and an Extended Kalman Filter (EKF) to provide drift-free, low-latency orientation outputs (Euler angles or quaternions) at up to 400 Hz. The LPMS-B2 has size at 39×39×8mm and weighs just 12g. It provides $\pm 0.01^\circ$ resolution, $\pm 0.5^\circ$ static accuracy, and $\pm 2^\circ$ RMS dynamic accuracy, with a battery life of more than six hours. As a result LPMS B2 is suitable for the experiments

Except the initial filtering of IMU sensor during the analysis, the dead-zone filtering apply on the data to get to provide a rapid and computationally lightweight filtering technique to minimize sensor noise while emphasizing relevant IMU-recorded head or neck movement data.

4.1.2 Questionnaires

The questionnaires presented to the participants will be clarified. As stated in the introductory section, psychological factors cannot be overlooked during these experiments. To begin with the pain assessment, the physiological examinations will encompass evaluations of pain intensity and a neck disability questionnaire. The 11-point Verbal Numerical Pain Scale (VNPS) will be used to objectively evaluate pain intensity, with participants requested to indicate their current pain level on it. The Verbal Numerical Pain Scale (VNPS) is well-known as an effective and dependable tool for evaluating personal pain perception. This scale spans the full spectrum of pain intensity, from complete absence of pain to the most severe pain conceivable, utilizing a scoring system from 0 to 10. A score of 0 indicates "no pain whatsoever," while a score of 10 represents "the most severe pain possible." Participants will be requested to evaluate the degree of pain

they are currently feeling using this scale. This assessment will yield a measurable sign of the person's personal pain level and will assist in assessing the effectiveness of treatment methods. Besides serving as a quick and easy instrument in clinical and scientific studies, the VNPS enables patients to communicate their feelings more accurately. Since individuals' views on pain differ based on numerous factors, utilizing this scale offers an approach that considers personal variations. Moreover, the VNPS will be employed not only to evaluate the present level of pain but also as an essential instrument for monitoring changes in pain over time. By conducting assessments before and after treatment, any rise or fall in the participant's discomfort will be recorded, and therapeutic strategies will be modified as required. Ultimately, using the VNPS will enhance the systematic and objective approach to pain assessment in both clinical settings and research, providing a consistent and reliable method[96][97]

Before the experiment, participants filled out the Neck Bournemouth Questionnaire (NBQ) to determine initial measurements of their neck symptoms and functional condition. The Neck Bournemouth Questionnaire (NBQ), recognized as an approved clinical assessment tool, assesses various facets of neck dysfunction across seven areas: pain severity, limitations in daily functionality, emotional distress, cognitive views on pain management, and hindrances to social involvement. Every item is rated on a scale from 0 to 10, offering a numerical assessment of the severity of neck pain and its psychosocial effects. This survey is especially pertinent for office employees, as it successfully reflects the complex aspects of work-related neck issues. By administering the NBQ prior to testing, the researchers gathered essential reference data to compare with later biomechanical measurements (like EMG and IMU recordings) and to evaluate any changes after experimental interventions. The NBQ's proven reliability and sensitivity to changes in clinical conditions render it an excellent tool for research and clinical assessment of neck pain, as shown in earlier studies[98]

Then, In this study, the validated Turkish version of the NBQ was utilized. It is the some the big reasons, why this questionnaire is selected.[99]This pre-experiment assessment guaranteed that participants' baseline pain and functional conditions were accurately recorded and factored into the final evaluation.

4.2 Experiments

In this part the steps of the experimental procedure explained. Firstly, the participants answer the NBQ and get informed about the experiment. Then, EMG sensors applied on the participant and MVC data get collected. As it mentioned on the previous parts, For UT participants get abducted and rotate the head for SCM. Therefore IMU sensor get connected to the participants forehead.

Therefore in the second phase of this experiment, the participant do 5 flexion 5 extension 5 right lateral flexion, 5 left lateral flexion, lastly, 5 right rotation and 5 left rotation Then, the participants do the same thing while they wearing the NACK device.

Lastly, in the third phase, the data taken from EMG and IMU sensors analyzed. Then the data taken from IMU import to the ADAMS and the simulations run with participants data.

4.2.1 Results

4.2.1.1 EMG Results

Figure 4.2 shows the flexion-extension (a) and lateral flexion (b) angles of the head over time for a representative participant, comparing both with and without the application of NACK. In the flexion-extension scenario, the peak angle without NACK is about $+64^\circ$, whereas with NACK it is restricted to around $+44^\circ$, indicating a decrease of about 31%. The minimum values indicate a decrease from 45° to 30° . Likewise, for lateral flexion, the highest angle reduces from $+40^\circ$ (without NACK) to $+20^\circ$ (with NACK), and

the lowest angle from 40° to 18°, leading to an overall decrease of about 50% in the range of motion. These reductions clearly.

Figure 4.1: Comparison of head angles with and without NACK a) head flexion-extension angle, and b) head lateral flexion angle

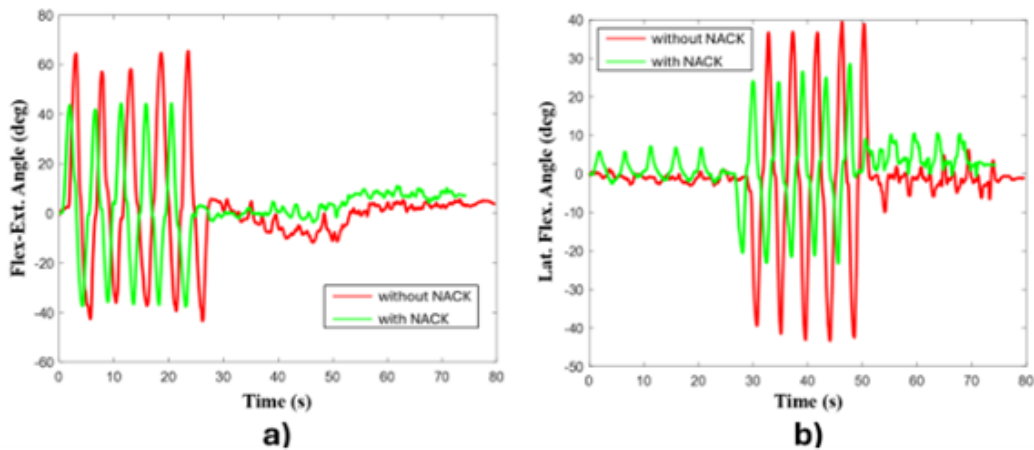


Figure 4.3 and 4.4 demonstrates muscle activation rates both with and without NACK. SCM-L shows the greatest activation, attaining around 0.65 without NACK and 0.4 with NACK, reflecting a significant decrease in muscle effort. A comparable pattern is seen for SCM-R, as activation drops from 0.45 to 0.3 with the device. In contrast, UT-R exhibits a rise in activation with NACK, increasing from 0.1 to 0.4. The activation of UT-L remains fairly stable, sustaining a constant level near 0.15 in both situations. These results indicate that although NACK successfully alleviates the burden on the SCM muscles, especially on the left side, it might transfer some of the load to the right UT.

Figure 4.2: Raw EMG Data for a) UT muscles (left), b) SCM muscles (left)

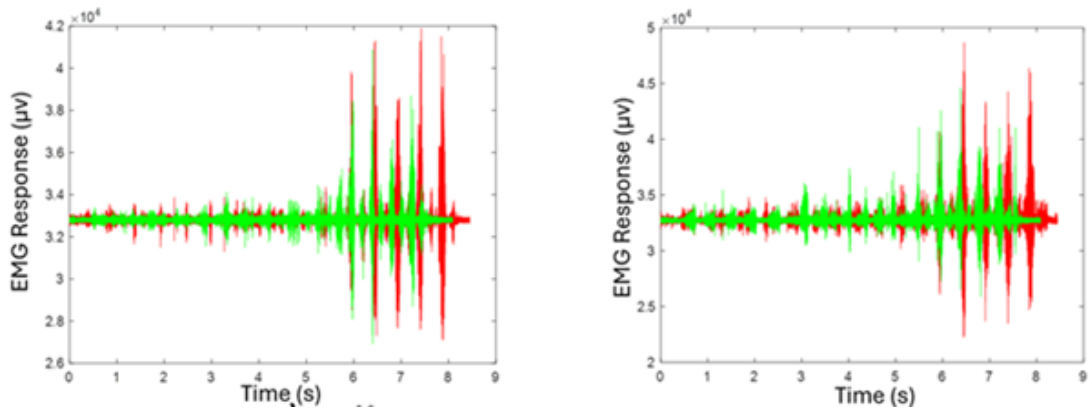
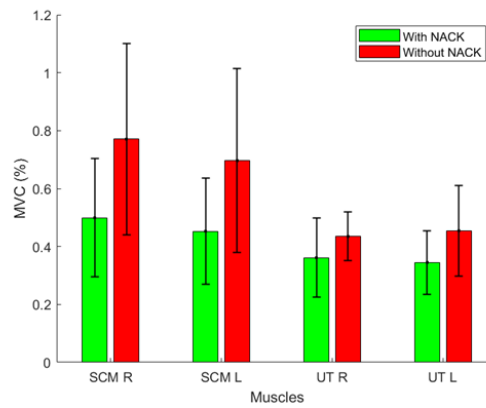


Figure 4.3: Muscle activations normalized by MVC



4.2.1.2 IMU RESULTS

For the Imu when we look the range of motion of the subjects in Figure 4.5 the system decrease all the participants RoM. Then in the Y-axis most decrease can be seen in terms of percent.,The angle of flexion-extension , angle of lateral flexion , and SCM muscle force from the ADAMS model (Fig. 4.6) were analyzed both with and without NACK. The angle of flexion-extension dropped from 65° to 40°, and the angle of lateral flexion declined from 40° to 28°. Furthermore, NACK lowered the maximum force applied by the SCM muscle from 65 N to 35 N. This decrease underscores the effectiveness of the

device in controlling muscular forces. These findings indicate that the assistive device effectively manages neck movements, maintaining a more neutral posture and alleviating strain.

Figure 4.4: a)RoM of Z-axis b)RoM of Y-axis c) Rom of X-axis

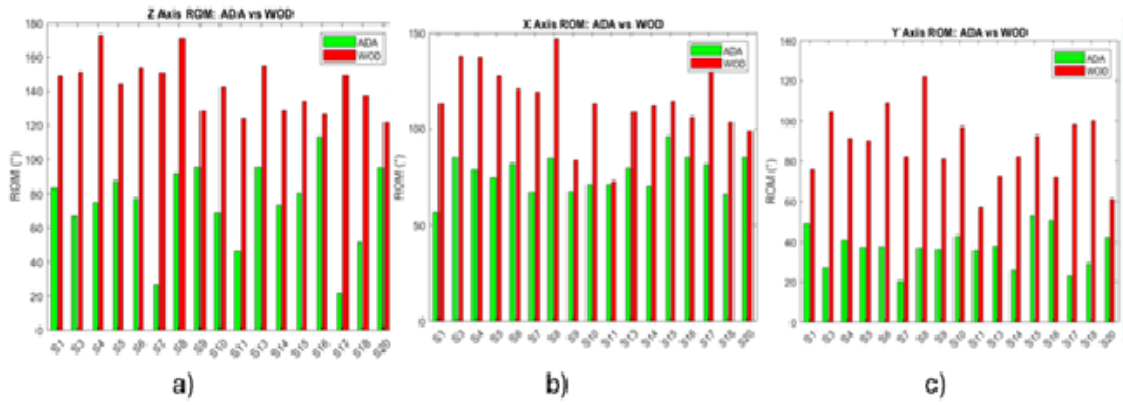
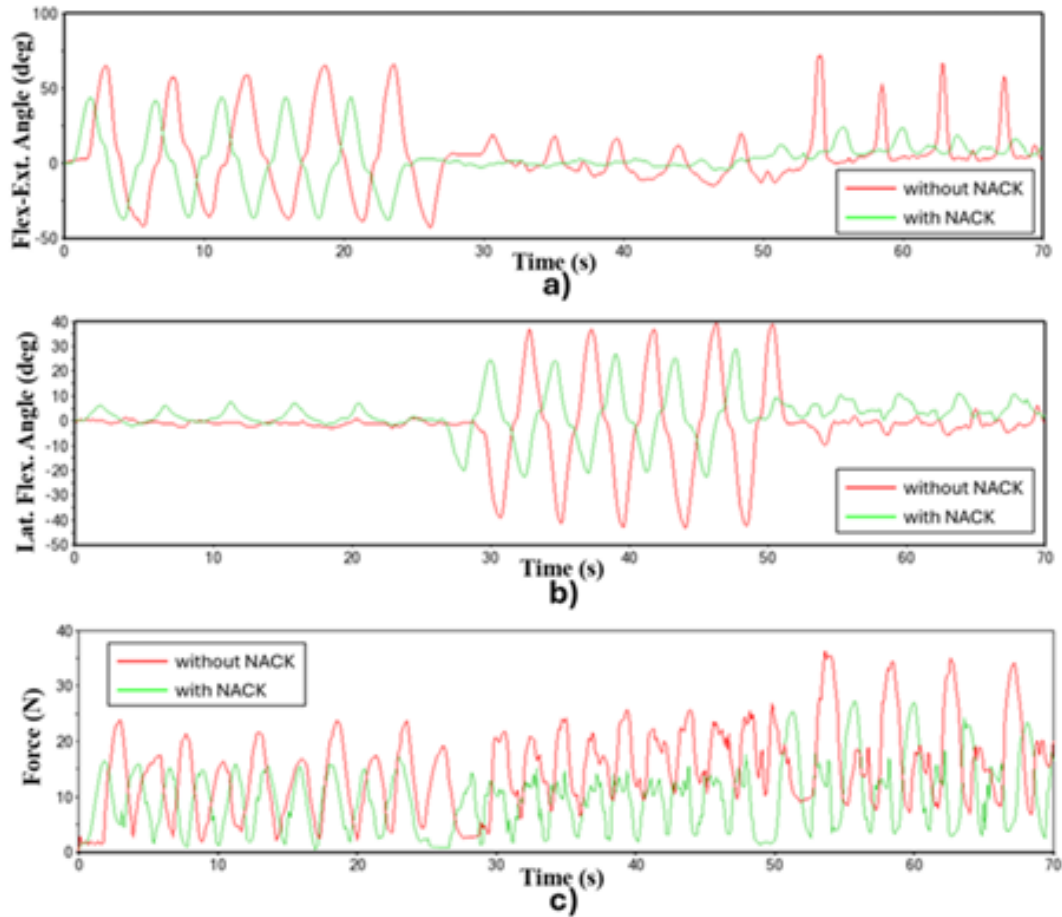


Figure 4.5: Comparison of the performance of NACK a) head flexion-extension angle, b) lateral flexion angle, and c) force of SCM muscle in ADAMS model



Ultimately, the longus colli, a deep flexor muscle in the cervical region, was examined using the ADAMS model, since EMG assessments are impractical for deep muscles. Table 4.1 displays the peak force, standard deviation, and force change range for one of the subjects, illustrating an example with and without the assistive device. The findings show that the NACK device suggested in this research lowers the peak force of deep muscles, evidenced by decreases in maximum force, standard deviation, and range throughout the whole simulation duration.

Table 4.1: *Comparison of force metrics with and without NACK*

Metric	Without NACK	With NACK	Improvement (%)
Maximum Force (N)	8.6	77	10.5
Standard Deviation (N)	0.78	0.36	53.8
Range of Force Change	5.2	35	32.7

4.2.1.3 Questionnaires Result

The findings of the Bournemouth Questionnaire, carried out before the experiments, are presented in Table 4.2. The values are derived from averaging the responses of 17 participants who have no known musculoskeletal conditions. The participants' average neck pain score of 2.5 (out of 10) also categorizes them as healthy.

Table 4.2: *Survey results of neck pain impact among participants*

Question	Average Answer
How severe has your neck pain been over the past week? (0 = No pain at all, 10 = Unbearable pain)	2.5
To what extent has your neck pain made it difficult for you to carry out daily tasks? (0 = Not at all difficult, 10 = I couldn't do any tasks)	1.5
To what extent has your neck pain affected your social or work life? (0 = Not at all, 10 = Completely prevented it)	1
How anxious or worried have you felt due to your neck pain over the past week? (0 = Not anxious at all, 10 = Extremely anxious)	0.6
How unhappy or depressed have you felt because of your neck pain over the past week? (0 = Not at all, 10 = Extremely depressed)	0.7
Have you avoided certain movements or situations due to your neck pain? (0 = Not at all, 10 = Constantly avoided)	1
How much control did you feel you had over your neck pain? (0 = No control at all, 10 = Complete control)	4.8

5. CONCLUSIONS

A comprehensive human head–neck model was effectively created and utilized in MSC ADAMS to aid in the design and validation of the NACK preventive neck orthosis. Through forward dynamics simulations, essential biomechanical metrics including head angular acceleration and angular velocity were evaluated to determine the model’s precision and realism. The analysis of these findings alongside current experimental data and ADAMS benchmarks showed that the created model accurately mimics real-world head–neck dynamics in standard office environments. These results validate the excellent accuracy of the MSC ADAMS head–neck model as a basis for creating, evaluating, and enhancing the NACK device. By precisely forecasting the neck’s reaction to various support setups, the design is essential for a preventive design of office workers. Furthermore, the model can be used for the biomechanical researches of neck.

After performing the , model analysis, user experiments, The modal analysis performed in this study shows that the created head–neck model can accurately forecast essential elements of human neck motion, such as range of motion, speed, and acceleration. This robust predictive ability established a firm basis for enhancing the mechanical design of the NACK device. Moreover, experimental trials and EMG readings verified that the device successfully diminishes muscle activation and balances force distribution throughout the neck muscles during operation. Collectively, these results confirm that the NACK system effectively simulates realistic neck biomechanics while successfully reducing muscle strain, emphasizing its promise as a viable preventive solution for those subjected to extended static positions.

This research showcases a proof-of-concept framework that is essentially based on a comprehensive, six degrees of freedom biomechanical representation of the human cervical spine. The sophisticated neck model maintains its complete predictive abilities, allowing precise simulation of kinematic outputs such as motion range, speed, and acceleration,

along with kinetic data like muscle activation rates and internal forces — incorporating the assessment of deep neck flexor muscles, which are vital for neck stability and are frequently overlooked in simpler models. This strong simulation framework directed the design and verification of the neck assistive concept, guaranteeing that the system efficiently minimizes excessive muscle activation and aids users in steering clear of dangerous ranges of motion and poor postures during extended sitting. Consequently, this design illustrates the functional importance of employing comprehensive biomechanical modeling to create a preventive neck orthosis, connecting sophisticated simulation with genuine ergonomic safety. As a result the contribution of this study can be summarized like this:

- A comprehensive head–neck multibody dynamics model has been created and verified through simulation results and experimental data from the literature.
- The outputs for the model’s motion range, speed, and acceleration are incorporated into a design framework to enhance the neck support system.
- The impact of the created support system on neck muscle activation and force distribution is measured through EMG recordings and IMU data evaluation during standard office tasks.
- The main design factors — such as the positioning and rigidity of the springs — are established to reduce unnecessary muscle engagement and avoid users adopting harmful neck positions during extended sitting.

Even though the method suggested in this research aims to enhance the main parameters of a theoretical neck support system, the technique is broad enough to be utilized in associated research areas later on. These supplementary areas are recognized as avenues for further research. The potential directions for future research arising from this study can be outlined as follows:

- Perform user studies and work alongside healthcare experts to enhance the NACK system's design and approach, drawing from actual user experiences to guarantee practicality, acceptance by users, and compliance with current medical and ergonomic standards.
- Explore the incorporation of cutting-edge materials, wearable tech, and intelligent sensing features into the neck support system to improve comfort and performance, while allowing for real-time feedback to facilitate personalized modifications.
- Assess the enduring impacts, financial viability, and expandability of the enhanced neck support system, considering strength, ecological consequences, production methods, and the likely market need for preventive ergonomic solutions.

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APPENDIX A. MATLAB CODES

In this part the IMU and EMG codes where using in experiments are shown

A.1 IMU Codes

```
clc
clear all
close all
clc

%% DATA IMPORT
filename = '15.05.2025_S18_ADC.csv';
D = readtable(filename);
% Extract base name without extension
[~, baseName, ~] = fileparts(filename);
% Optional: Split into parts if needed
parts = split(baseName, '_'); % {"15.04.25", "S1", "WOD"}

% Extract session and condition names
sessionName = parts{2}; % "S1"
conditionName = parts{3}; % "WOD"

Mod_D=[];
X_Acc=D(:,4);
Y_Acc=D(:,5);
Z_Acc=D(:,6);
X_Rot=D(:,13);
Y_Rot=D(:,14);
Z_Rot=D(:,15);
Time=D(:,2);
X_Acc=table2array(X_Acc); %g
Y_Acc=table2array(Y_Acc); %g
Z_Acc=table2array(Z_Acc); %g
X_Rot=table2array(X_Rot); %g
Y_Rot=table2array(Y_Rot); %g
Z_Rot=table2array(Z_Rot); %g

% Subtract 2nd row values from all rows
X_Rot = X_Rot - X_Rot(2);
```

```

Y_Rot = Y_Rot - Y_Rot(2);
Z_Rot = Z_Rot - Z_Rot(2);
X_Acc_Mod=[];

for i=1:1:length(X_Acc)
    if X_Acc (i) <= 0.009 & X_Acc (i) >= -0.015;
        X_Acc_Mod (i) = 0;
    else
        X_Acc_Mod (i) = X_Acc (i);
    end
end

Y_Acc_Mod=[];

for i=1:1:length(Y_Acc)
    if Y_Acc (i) <= -0.02 & Y_Acc (i) >= -0.025;
        Y_Acc_Mod (i) = 0;
    else
        Y_Acc_Mod (i) = Y_Acc (i);
    end
end

Z_Acc_Mod=[];

for i=1:1:length(Z_Acc)
    if Z_Acc (i) <= 0.009 & Z_Acc (i) >= -0.015;
        Z_Acc_Mod (i) = 0;
    else
        Z_Acc_Mod (i) = Z_Acc (i);
    end
end

Time=table2array(Time); %Sec

Mod_D=[Time,X_Acc_Mod',Y_Acc_Mod',Z_Acc_Mod'];
D=[Time,X_Acc,Y_Acc,Z_Acc,X_Rot,Y_Rot,Z_Rot];

Mod_Z=[Time,Z_Acc_Mod'];
Mod_Y=[Time,Y_Acc_Mod'];
Mod_X=[Time,X_Acc_Mod'];

```

```

Z=[Time,Z_Acc];
Y=[Time,Y_Acc];
X=[Time,X_Acc];
Z_R=[Time,Z_Rot];
Y_R=[Time,Y_Rot];
X_R=[Time,X_Rot];

dt=Time(length(Time))-Time(length(Time)-1)

% Generate output filenames dynamically
filenameZ = sprintf('%s_%s_Z_Rot_3.txt', sessionName, conditionName);
filenameY = sprintf('%s_%s_Y_Rot_3.txt', sessionName, conditionName);
filenameX = sprintf('%s_%s_X_Rot_3.txt', sessionName, conditionName);

% Assuming X_R, Y_R, Z_R already computed
writematrix(Z_R, filenameZ);
writematrix(Y_R, filenameY);
writematrix(X_R, filenameX);

%% Plot
figure(1)
plot(Time,X_Acc,'b','LineWidth',3)
hold on
plot(Time,Y_Acc,'g','LineWidth',3)
plot(Time,Z_Acc,'r','LineWidth',3)
grid on
xlabel('Time')
ylabel('Acc (g)')
% title('Time & Acc')
legend('X_Acc','Y_Acc','Z_Acc')

set(gcf, 'Color', 'white');
% Set axes background to white
set(gca, 'Color', 'white');
grid off;
hold off;

figure(2)
plot(Time,X_Rot,'b','LineWidth',3)
hold on

```

```

plot(Time,Y_Rot,'g','LineWidth',3)
plot(Time,Z_Rot,'r','LineWidth',3)
grid on
xlabel('Time')
ylabel('Rot (Deg)')
% title('Time & Rot')
legend('X_Rot','Y_Rot','Z_Rot')

set(gcf, 'Color', 'white');
% Set axes background to white
set(gca, 'Color', 'white');
grid off;
hold off;
figure(3)
plot(Time,X_Acc,'b','LineWidth',3)
hold on
plot(Time,Y_Acc,'g','LineWidth',3)
plot(Time,Z_Acc,'r','LineWidth',3)
plot(Time,X_Rot,'b.','LineWidth',3)
plot(Time,Y_Rot,'g.','LineWidth',3)
plot(Time,Z_Rot,'r.','LineWidth',3)
grid on
xlabel('Time')
ylabel('Acc & Rot')
title('Time & Acc & Rot')
legend('X_Acc','Y_Acc','Z_Acc','X_Rot','Y_Rot','Z_Rot')

clc;
clear;
close all;

subjects = {'S1','S3','S4','S5','S6','S7','S8','S9','S10','S11','S12'};

% Initialize storage
rom_ada_x = [];
rom_wod_x = [];
rom_ada_y = [];
rom_wod_y = [];
rom_ada_z = [];
rom_wod_z = [];

for i = 1:length(subjects)
    subj = subjects{i};

```

```

% === Load TXT files ===
ada_x = load(sprintf('%s_ADA_X_Rot_3.txt', subj));
wod_x = load(sprintf('%s_WOD_X_Rot_3.txt', subj));
ada_y = load(sprintf('%s_ADA_Y_Rot_3.txt', subj));
wod_y = load(sprintf('%s_WOD_Y_Rot_3.txt', subj));
ada_z = load(sprintf('%s_ADA_Z_Rot_3.txt', subj));
wod_z = load(sprintf('%s_WOD_Z_Rot_3.txt', subj));

% === Get ROM (rotation values are in column 2) ===
rom_ada_x(end+1) = max(ada_x(:,2)) - min(ada_x(:,2));
rom_wod_x(end+1) = max(wod_x(:,2)) - min(wod_x(:,2));
rom_ada_y(end+1) = max(ada_y(:,2)) - min(ada_y(:,2));
rom_wod_y(end+1) = max(wod_y(:,2)) - min(wod_y(:,2));
rom_ada_z(end+1) = max(ada_z(:,2)) - min(ada_z(:,2));
rom_wod_z(end+1) = max(wod_z(:,2)) - min(wod_z(:,2));
end

% === Plot X-axis Comparison ===
figure;
bar([rom_ada_x' rom_wod_x'], 'grouped');
title('X Axis ROM: ADA vs WOD');
ylabel('ROM (°)');
xticks(1:length(subjects));
xticklabels(subjects);
legend('ADA', 'WOD');
set(gcf, 'Color', 'white');
set(gca, 'Color', 'white');
grid off;

% === Plot Y-axis Comparison ===
figure;
bar([rom_ada_y' rom_wod_y'], 'grouped');
title('Y Axis ROM: ADA vs WOD');
ylabel('ROM (°)');
xticks(1:length(subjects));
xticklabels(subjects);
legend('ADA', 'WOD');
set(gcf, 'Color', 'white');
set(gca, 'Color', 'white');
grid off;

% === Plot Z-axis Comparison ===

```

```

figure;
bar([rom_ada_z' rom_wod_z'], 'grouped');
title('Z Axis ROM: ADA vs WOD');
ylabel('ROM (°)');
xticks(1:length(subjects));
xticklabels(subjects);
legend('ADA', 'WOD');
set(gcf, 'Color', 'white');
set(gca, 'Color', 'white');
grid off;

```

A.2 EMG Codes

```

clc; clear;

mvc = readmatrix('15.04.2025_S4_MVC.txt');
wod = readmatrix('15.04.2025_S4_WOD.txt');
ada = readmatrix('15.04.2025_S4_ADA.txt');
adc = readmatrix('15.04.2025_S4_ADC.txt');

% Sampling frequency
fs = 1000;

mvc_data = mvc(:, 3:6);
wod_data = wod(:, 3:6);
ada_data = ada(:, 3:6);
adc_data = adc(:, 3:6);

muscles = {'SCM_R', 'SCM_L', 'UT_R', 'UT_L'};
conditions = {'WOD', 'ADA', 'ADC'};
data_struct = {wod_data, ada_data, adc_data};

[b_bp, a_bp] = butter(4, [20 490]/(fs/2), 'bandpass');
[b_lp, a_lp] = butter(4, 6/(fs/2), 'low');

results = [];

for m = 1:length(muscles)
    mvc_max = max(mvc_data(:, m));
    for c = 1:length(conditions)
        raw = data_struct{c}(:, m);
        filtered = filtfilt(b_bp, a_bp, raw);           % bandpass
        rectified = abs(filtered);                     % rectify
    end
end

```

```

        envelope = filtfilt(b_lp, a_lp, rectified); % envelope
        percent_mvc = envelope / mvc_max * 100;
        avg_val = mean(percent_mvc);
        results = [results; {'S4'}, muscles{m}, conditions{c}, avg_v
    end
end

T = cell2table(results, 'VariableNames', {'Subject', 'Muscle', 'Condi
writetable(T, 'S4_Filtered_Mean_MVC.csv');
disp(T);

% Clear workspace and close all figures
clc;
close all;
clear;

% Load the data files
load S4_MVC.mat; % Contains Maximum Voluntary Contractions
load S4_AD.mat; % Data with the device
load S4_WOD.mat; % Data without the device

% Sampling frequency
fs = 1000;

% Cutoff frequencies for filtering
fLow = 20; % Low cutoff for bandpass
fHigh = 490; % High cutoff for bandpass
fEnvelope = 6; % Low-pass cutoff for envelope

% Convert tables to arrays if needed
MVC = table2array(S4MVC);
AD = table2array(S4AD);
WOD = table2array(S4WOD);

% Muscle columns (3 to 6 represent the four muscles)
muscles = 3:6;

% Initialize arrays for normalized activations
normalized_AD = zeros(1, length(muscles));
normalized_WOD = zeros(1, length(muscles));

% Loop through each muscle
for muscle_idx = 1:length(muscles)

```

```

muscle_col = muscles(muscle_idx); % Column index for the muscle

% Extract raw EMG data for current muscle
emg_AD = AD(:, muscle_col);
emg_WOD = WOD(:, muscle_col);
mvc_value = max(MVC(:, muscle_col)); % Get MVC for normalization

% Bandpass filter the raw EMG signals
[b, a] = butter(4, [fLow, fHigh] / (fs/2), 'bandpass');
emg_AD_filtered = filtfilt(b, a, emg_AD);
emg_WOD_filtered = filtfilt(b, a, emg_WOD);

% Rectify the signals
emg_AD_rectified = abs(emg_AD_filtered);
emg_WOD_rectified = abs(emg_WOD_filtered);

% Low-pass filter the rectified signals to get the envelope
[bEnv, aEnv] = butter(4, fEnvelope / (fs/2), 'low');
emg_AD_envelope = filtfilt(bEnv, aEnv, emg_AD_rectified);
emg_WOD_envelope = filtfilt(bEnv, aEnv, emg_WOD_rectified);

% Normalize by MVC
normalized_AD(muscle_idx) = mean(emg_AD_envelope) / mvc_value * 1
normalized_WOD(muscle_idx) = mean(emg_WOD_envelope) / mvc_value * 1
end

% Define muscle names for plotting
muscle_names = {'SCM R', 'SCM L', 'UT R', 'UT L'};

% Create bar plot with specific colors
figure;
b = bar([normalized_AD; normalized_WOD]', 'grouped');

% Set colors: red for 'With Device' and blue for 'Without Device'
b(1).FaceColor = 'r'; % Red color for 'With Device'
b(2).FaceColor = 'b'; % Blue color for 'Without Device'

% Set x-axis labels and other plot details
set(gca, 'XTickLabel', muscle_names);
xlabel('Muscles');
ylabel('MVC (%)');
legend('With Device', 'Without Device');
title('');

```

grid off;



