

DESIGN OF BALANSENS: SENSORY, ANTERIOR AND FEEDBACK BALANCE ANALYSIS, REHABILITATION, AND SUPPORT ROBOTIC PLATFORM

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DESIGN OF BALANSENS: SENSORY, ANTERIOR AND FEEDBACK BALANCE ANALYSIS, REHABILITATION, AND SUPPORT ROBOTIC PLATFORM

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To My Family.

ABSTRACT

In the thesis, the design and development evaluation of BalanSENS toward the realization of the Integrated Balance Rehabilitation (I-BaR) framework is presented. BalanSENS is designed to encourage active participation by integrating multi-sensory information with the co-improvement of sensory and motor functions. Moreover, it can offer individual rehabilitation design with the integration of three phases. The first phase provides foot-ankle muscle activation and movement sensation development. In the second phase, sensory weighting skills and upper extremities independence can be improved by using multi-sensory input. In the last/stepping phase, walking parameters are aimed to be improved with modulated distance. The parallel manipulator is designed through simulations to determine actuation properties and analyze the load-bearing capacity and feasibility of the materials. Drawing from simulation outcomes, an operational 3 DoF platform is constructed to demonstrate their design suitability for the I-BaR framework. Furthermore, design evaluations demonstrated promising results in quantifying force and real-time data monitoring during the passive ankle preparation phase.

ÖZETÇE

Bu tezde, BalanSENS'in Entegre Denge Rehabilitasyonu (I-BaR) programının gerçekleştirilmesine yönelik tasarım ve geliştirme değerlendirmesi sunulmaktadır. BalanSENS, çoklu duyuşal bilgileri duyuşal ve motor fonksiyonların birlikte geliştirilmesiyle bütünleş-tirerek aktif katılımı teşvik etmek için tasarlanmıştır. Ayrıca üç aşamanın ente-grasyonu ile bireysel rehabilitasyon tasarımı da sunabilmektedir. İlk aşama ayak-ayak bileği kaslarının aktivasyonunu ve hareket duyuşunun gelişimini sağlar. İkinci aşamada çoklu duyuşal girdi kullanılarak duyuşal ağırlıklandırma becerileri ve üst ekstremité bağımsızlığı geliştirilebilir. Son olarak, adım aşamasında modüle edilmiş mesafe ile yürüme parametrelerinin iyileştirilmesi amaçlanmaktadır. Paralel ma-nipülatörün, malzemelerin özelliklerini ve fizibilitesinin analizleri için simülasyonlar kullanılmıştır. Simülasyon sonuçlarından yola çıkarak, I-BaR programı için tasarım uygunluğunu göstermek amacıyla operasyonel bir 3 DoF paralel manipülatör pro-totipi yapıldı. Ayrıca pasif ayak bileği hazırlık değerlendirmeleri sonucunda sistem, kuvvetin ölçülmesinde ve gerçek zamanlı veri izlemede umut verici sonuçlar ortaya koydu.

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List of Acronyms

AAN	Assist As Needed
ADL	activities of daily living
APA	anticipatory postural adjustments
CNS	central nervous system
COM	center of mass
COP	center of pressure
CPA	compensatory postural adjustments
CSRT	Choice Stepping Reaction Time
CTC	computed torque control
DoF	Degree of Freedom
EMG	electromyography
EtherCAT	Ethernet for Control Automation Technology
FRA	Foot Rotation Angle
GRF	ground reaction force
I-BaR	Integrated Balance Rehabilitation
IC	Integrated circuit
IMU	Inertial Measurement Unit
MoI	moment of Inertia
MS	Multiple Sclerosis
PBT	perturbation-based balance training
PD	Parkinson's disease
PNS	peripheral nervous system
QoL	Quality of Life
RMSE	Root-Mean-Square Error
ROM	range of motion

VF vibrotactile feedback

VR virtual reality



CHAPTER I

INTRODUCTION

Neurological disorders impact approximately 1 billion individuals worldwide, representing a diverse range of socio-economic statuses, age groups, and ethnicities. Furthermore, an estimated 6.8 million people die each year as a result of neurological diseases [1, 2, 3]. The number of deaths from neurological disorders and disability has increased dramatically in the last 30 years, particularly in low- and middle-income countries; a further increase is foreseen worldwide due to growth in population and aging. They are predicted to impact about 10% of the global burden of disease and 25% of years lived with disability [1]. Along with the cost of care for neurological disorders, the financial loss related to unemployment, etc., was estimated to be \$9 billion per year in Canada in 2007 [4], and about €1135 million in the United Kingdom in 2010 [2]. In the United States, it was estimated that expenses only for stroke alone exceeded \$33.6 billion in 2015 [3]. In order to provide adequate service in this regard, it is vital to take action to meet the growing demand.

Individuals with neurological disorders experience walking and movement restrictions, balance, motor control, exhaustion, and other health problems, which all directly impact their quality of life (QoL) due to impairments in the central and peripheral nervous systems (CNS and PNS). The nature and severity of the condition may differ from patient to patient depending on the lesional region at the affected CNS and PNS. In Multiple Sclerosis (MS), commands are conveyed through the nerve slowly or cannot be transmitted due to degeneration of the myelin sheath [5, 6]. In Parkinson's disease (PD), depigmentation of the substantia nigra and damage to dopamine-producing cells result in deficiencies in the balance control [7]. Further, in

stroke disease, focal neurological function losses occur due to infarction or bleeding in the relevant part of the brain [8]. The CNS integrates information from the visual, vestibular, proprioceptive, and cognitive systems, and PNS through continuous sensory re-weighting; provides postural control in static and dynamic situations [9]. This integration of multi-sensory information is disrupted in neurological diseases, causing balance deficits and an increase in fall rate [10]. More than 75% of MS patients have symptoms of poor balance. Further, approximately 60% of MS patients reported at least one fall in the previous three months, and more than 80% reported impairments in activities of daily living (ADL) [11, 12]. It has been reported that approximately 73% of stroke patients and 45-68% of Parkinson's patients experience at least one fall per year [13]. The factors mentioned above lead to a sedentary lifestyle and accordingly deteriorate patients' social health, which is a subdomain of QoL. This lifestyle may also lead to additional health-related issues such as obesity, diabetes, and heart disease [12, 14], which further dramatically deteriorates their QoL [15]. Due to the factors mentioned above, the search for improvements in existing rehabilitation methods continues.

The somatosensory function includes the senses of touch, vibration, pressure, proprioception, pain, and temperature. Impairments in this function negatively affect the ability to perceive, distinguish, and recognize the senses in the body [16]. Consequently, disorientation accompanied by abnormal movements, balance disorders, muscle weakness, and inability to maintain postural control (stabilize the body against gravity and perturbation) may occur [17]. For instance, it is reported that post-stroke individuals experience high rates of somatosensory impairment, ranging between 65-85% [18]. A research with self-questionnaire demonstrated that individuals with somatosensory and motor impairments suffer from lower walking capacities and lower levels of independence in ADL [19, 20, 21]. Since restoring walking ability is a primary objective for many stroke patients, establishing the best treatments for balance, gait,

and mobility were identified as one of the top ten stroke research priorities [22]. In a recent survey conducted with 145 stroke individuals, 43% of individuals reported decreased sensation in their feet; sensory impairment was indicated to be the second most common foot problem after the loss of strength [23, 24]. Limitations in walking, high fall rate, and impairments in foot-ground contact and sense of foot position sense and hence, decrease the outdoor activities in the community.

This thesis proposes a novel Integrated Balance Rehabilitation (I-BaR) framework and BalanSENS robotic platform. The I-BaR framework is explained in three phases: ankle-foot, balance, and stepping. The ankle-foot/preparation phase aims to increase mobility, joint proprioception, and sole sensation. The balance phase aims to enhance sensory weighting abilities via perturbation-based balance training (PBT). Lastly, the stepping phase aims to enhance walking characteristics by performing stepping at adjustable-distance target points with internal/external perturbation. BalanSENS robotic platform is designed to be suitable for the I-BaR framework. In the first phase, the patient is seated in a chair while the ankle range of motion (ROM), strength, proprioception, and sensory training are performed by the 3-DoF parallel manipulator. The control architecture of the parallel manipulator is developed to be suitable to the assist as needed (AAN) paradigm so the patient's active participation level can be adjusted. In the second phase, PBT is applied using the same 3-DoF parallel manipulator. Also, the support surface on the end effector of the parallel manipulator can be adjusted between 0-20°, so that the difficulty of treatment can be changed during PBT. In addition, as in the ankle phase, the AAN paradigm and increasing perturbation level will be used in this phase. In the last stage, according to the given signal, the patients will be asked to step to the target points that are placed straight and diagonally with adjustable distance. The BalanSENS includes different sensors, e.g., vibration motors, load cells, Inertial Measurement Unit (IMU), and electromyography (EMG), so that throughout all phases, the measurements can

be calculated quantitatively, the progress of the patients can be evaluated objectively, and multi-modal feedback can be given.

1.1 *Balance Control Mechanisms*

Balance control, is highly activity-specific and falls into three categories:

- *Static/dynamic steady-state balance control*: is sustaining a stable posture while sitting, standing, or walking.
- *Proactive balance control*: is activated before the predicted perturbation. The CNS uses postural adjustments, i.e., anticipatory and compensatory postural adjustments (APAs/CPAs), envisaged as a muscular adjustment mechanism to provide balance control while maintaining body balance and vertical posture during different conditions. These adjustments engage and activate the trunk and lower extremity postural muscles before an impending external/internal perturbation occurs. It reduces the risk of balance deterioration by regulating the body's center of mass (CoM) position.
- *Reactive balance control*: is activated after the perturbation to compensate CoM deviation [25, 26]. The CNS uses CPAs as a muscular adjustment mechanism to provide this balance control. These adjustments are triggered by the sensory control signal and allow the CoM to be repositioned once it is disturbed [27].

After people lose their balance, they only have a few seconds to coordinate and stabilize their posture [9, 28]. Postural perturbations such as sliding and tripping in everyday situations vary widely and are highly unpredictable. Recently, it has been proposed that studying the processes of APAs and CPAs may reveal vital information about postural control and falls. Numerous research findings show that due to postural deficiencies, falls occur during ADL [29, 30, 31]. The duration and magnitude of muscular activation were measured in different studies, and significant APAs

deficiencies were identified. There is evidence that improvements in the production of APAs can be achieved even after a single training session in individuals with stroke [28]. In another study [32] with elderly healthy individuals shows that APAs improvement can also be achieved after four weeks of external perturbation training. These studies show that individuals exposed to predicted perturbations provide better compensatory activity in the muscles (improvement in APAs and CPAs) and more adequate body pressure center changes with the use and production of strong APAs [28, 32]. After training, early muscle activation and reduced CoM excursions took place, which is substantial evidence that retraining of APAs is possible. These results form an essential basis for investigating training effectiveness focused on improving long-term APAs, CPAs, and reaction time in increasing individuals' postural control. Furthermore, they provide a background for the development of perturbation programs to improve postural control and balance, and prevent falls.

Reaction time is commonly considered in clinical diagnosis, treatment, and follow-up stages to determine the severity of postural control deficiency (APAs and CPAs) in somatosensory-based motor and neurological disorders [33, 34]. It is the time elapsed between the onset of a stimulus and when the patient's response to that stimulus begins, which is physiologically divided into five parts. These are (1) seeing the stimulus at the receptor level, (2) transmitting the stimulus to the CNS, (3) transmitting the stimulus through the nerves, (4) generating the effector signal, (5) transporting the signal to the muscles through the CNS for the mechanical work to be done [35, 36, 37]. When they examined the surface EMG activity of the lower extremity muscles and the center of pressure (CoP) of both falling and non-falling stroke patients, it was found that the falling group had lower muscular electrical activity. Therefore, they need longer reaction times to prepare the posture and initiate movements through APAs [38].

The reactive and proactive balance controls are managed by the activation of

different neurological mechanisms of the CNS. The impairment in one of the postural adjustments (APAs and CPAs) and their ability to affect each other negatively highlights the importance of assessing and training them in rehabilitation. The loss of balance may occur due to an unpredictable external force or failure of balance control after external/internal perturbation, i.e., fast and voluntary movement of the extremities. Therefore, it is possible to observe improvements in both postural adjustments (proactive and reactive balance control) with effective fall prevention training [39, 40, 41].

A novel approach to analyzing APAs employs Fitts' law to explain the relationship between APAs parameters, the length of the step, and the size of the stepping target. It is a valuable method for various target-directed movements and quantifying stability control factors. Thus, previous literature on able-bodied individuals report that Fitts' law is a valid way to explain the time to complete the foot-reaching task and APAs levels [42, 43, 44, 45]. A voluntary step initiation is a self-perturbation of balance with a change in the base of support and the transition from a static to a dynamic posture, so the velocity and accuracy of movement can be assessed with Fitts' tasks since coordinated muscle activation prior to voluntary movement is utilized to maintain the posture. The Choice Stepping Reaction Time (CSRT) test is a simple activity that evaluates a person's ability to immediately trigger and execute a step with Fitts' law. The subject must step on one of the numerous targets put in front of or around them as rapidly as feasible. The time it takes to attain the goals, i.e., reaction time, is a promising strategy for assessing fall risk among the elderly population since they have a significantly longer duration in reaction time than non-fallers. Furthermore, a few studies investigated the relationship between speed and accuracy control in foot-reaching tasks. Patients were asked to use their affected leg to step to targets with different sizes and at varying distances during these tasks [36, 40, 46]. However, these target points only include a switch button to calculate

the reaction time. In other words, they cannot measure ground reaction force (GRF) and give haptic feedback to the patient. Yet, as neurological patients report decreased sensation on the sole and trouble in weight shifting, so it is vital to measure GRF for appropriate feedback and address aforementioned challenges in the training [47].

1.2 The Perturbation-Based Balance Training

The PBT is a type of exercise in which participants are intentionally disturbed to improve reactive balance reactions by training the individual neuromuscular responses [48, 49, 50]. This training requires performing rapidly occurring sequential whole-body movements and applying large and sudden disruptive forces to stabilize CoM [51]. With the development of balance reactions, an increase in the ability to respond to the loss of balance in unpredictable ADL and, consequently, a decrease in fall rate can be achieved [48].

In addition to the physiotherapist's manual pushes and pulls (lean and release test) in PBT studies, treadmill acceleration-deceleration, and inclined/moving platforms have been implemented in recent years to mimic external perturbations in daily life [51, 52, 53]. It is suggested that perturbation training while walking could effectively minimize fall rates in elderly people [54, 55]. Current research suggests that CPAs in the elderly can be improved using PBT and that these improvements can be sustained for up to a full year after training [54]. Another study shows that only a single session of perturbation is sufficient to provide permanent improvements in reactive balance control and prevent falls in elderly individuals [28]. However, to the best of our knowledge, no study has been conducted on the optimal dosage of perturbation training to induce permanent changes in reactive balance control.

Although the current PBT is an approach to decrease the fall rate, it is still far from the realistic condition. The limited type of perturbations performed with existing devices and techniques may reduce individuals' capacity to adapt and generalize

the effects of PBT training to ADL. On top of that, different perturbation modalities in the PBT programs can be considered highly important to train balance reactions to match with a variety of situations and motor tasks [56, 57, 58].

1.3 Proprioceptive/Sensorimotor Training

Sensory-motor training gradually improves an individual's ability to re-weight and integrate sensory inputs in order to control balance and prevent falls in different somatosensory input situations [59]. According to these approaches, new technological devices and paradigms are being developed to promote neurorehabilitation from the CNS to the PNS. Moreover, the participation of cognitive functions increased by integrating multi-sensory feedback thus, it improved rehabilitation effectiveness [60, 61, 62].

Smania et al. [63] show significant improvements in the ability of stroke patients to maintain balance control with a unique training program based on weight transfer and balance exercises performed under different manipulation of sensory inputs. Derakhshanfar et al. [64] report that exteroceptive and proprioceptive stimulations, which include sensory intervention, are effective in improving motor function and ADL. These studies show that the neuromotor system can be activated correctly by providing a change in the sensory inputs to muscle and joint receptors as well as the skin receptors of patients' feet [65]. Lim [66] reports that a multi-sensory training program significantly improves proprioception and balance ability in patients; however, these types of studies are very limited.

The main goal of rehabilitation is the recovery of lost motor skills permanently and as quickly as possible. An effective way of training in motor learning can go through the optimization of given tasks and feedback by variable sensory inputs such as pressure, vibration, and proprioception. These inputs not only facilitate motor learning but also develop compensatory mechanisms and strategies to overcome the

loss of motor function resulting from a damaged neuromuscular system [67]. Feedback (visual, auditory, or tactile) helps to improve complex motor learning. However, multi-modal stimuli are more dominant in daily life than uni-modal stimuli since they provide convenience in ADL. Healthy individuals process stimuli in different modalities simultaneously. Multi-modal stimuli enable the learning of several aspects of a movement at the same time. Certain advantages of each modality are exploited, such as the ability of visualizations to show spatial aspects and audio or tactile feedback to show temporal aspects. Moreover, it is reported that this sensory enhancement facilitates the transition between the senses (using other senses when one sense is inadequate). The researchers hypothesize that after training with multi-modal stimuli, sense processing would be active even when only uni-modal stimuli are present. For example, the learning process of motion detection tasks progressed positively even when auditory feedback was canceled after training with audio-visual feedback. This shows the importance of multi-modal training for complex motion recovery even at a further level [67, 68, 69].

In recent years, rehabilitation strategies that include the active participation of patients and task-oriented exercises during rehabilitation sessions are carried out by virtual reality (VR). It is integrated to increase the attention and motivation of individuals with the desire for reward and success by giving continuous feedback [70]. There are two different approaches to using VR in rehabilitation. The first one is called serious games, and they are specifically designed for the rehabilitation robot/method. The second one is called exercise games and refers to the use of games that already exist for entertainment purposes in rehabilitation [71]. With the integration of robotic rehabilitation in the VR environment, it is aimed to provide visual feedback about the results of the movements performed during the rehabilitation process, increase awareness about the quality of the movements, and accelerate motor learning [72]. By including haptic cues during these applications, gait parameters

such as symmetry, balance, and muscle activation patterns can be improved collectively. Thus, it is observed that the motor and cognitive status of patients can be improved at high rates by providing multi-sensory feedback and repetition of tasks with the robotic device, encouraging patients to actively participate in rehabilitation [72, 73].

A systematic review and meta-analysis are conducted to investigate the effectiveness of vibrotactile feedback (VF) on balance and gait rehabilitation. It shows that haptic feedback could represent a helpful intervention [13] which is generally divided into two types: kinesthetic and tactile. The former cue usually contains a sense of force and provides the user with a spatial frame of reference [61, 74], while the latter cue usually includes a sense of vibration, texture, or pressure. Such feedback can be delivered via existing interfaces, which offer to the user kinesthetic and tactile sensations [75, 76]. When paired with other feedback, i.e., kinesthetic [77] with visual [78], the beneficial effects of VF on gait and balance parameters are found to be stronger. Because of these characteristics, VF can be employed as a supportive sensory stimulus in the context of a rehabilitation intervention focusing on sensorimotor integration. VF is also shown to be therapeutic for patients with neurological disorders [79, 80, 81].

The interdependence of sensory, cognitive, and motor processes, as well as the need for integrated training, are increasingly employed in therapies to improve ADL, motor skills, and balance, in the meantime, to decrease fear of falling. In other words, APAs and CPAs (proactive and reactive control adjustments) can be retrained to improve QoL.

1.4 Robot-Aided Rehabilitation

Robot-assisted rehabilitation is continuously gaining prominence as it provides more efficient training and objective evaluation than conventional rehabilitation procedures

[82, 83, 84]. Furthermore, conventional methods require at least three physiotherapists to support the lower extremity and trunk of the patient manually. Another limitation is that the effectiveness of the rehabilitation during these practices depends on the personal knowledge and experience of the therapist. It is reported that the demand for physiotherapists is increasing continuously to match the number of patients due to the increase in the aging population worldwide. The use of robotic devices to address these challenges is encouraged to shift the adoption of rehabilitation clinics from conventional methods to robotic-assisted rehabilitation. Hence, high-quality therapy sessions can be achieved at a relatively low cost and with significantly less effort [85, 86, 87, 88].

Most neurological patients have a reduced ROM in their ankle as well as muscle strength in their body. In the literature, muscle-strengthening for ankle rehabilitation has been considered with three main phases: ROM, strength, and proprioceptive training, respectively, and they require different control methods. At the beginning of the rehabilitation, the patients have limited ankle mobility; thus, passive ROM exercises are required (the patient is passive, and the device is fully active) so that ankle ROM can be recovered via full assistance of the device. Then, active ROM exercise is utilized by reducing the level of assistance from the device, and the patient should put effort into initiating the motion against the device. The second phase is about improving the ankle stability; as the patient's response improves, the resistance level of these strengthening activities increases. In most cases, they are unable to exert enough force to complete the exercises, so the robot should assist them. In the final phase, proprioceptive training should be performed to improve the balance control [89, 90, 91]. The control strategies in the stated phases are illustrated in Table 1.

Even though, robotic-assisted rehabilitation has a promising effect on the treatment itself, especially with the capability of being repetitive and task-specific, implementation of these potentials at the desired level is still challenging. There are

Table 1: Control Methods in Rehabilitation

Exercise method	Patient mode	Control method
ROM	Passive	Position Control
	Active	Assistive Control
Strength Training	Active- Isometric	Position Control
	Active- Isotonic	Admittance Control
Proprioceptive Training	Active	Hybrid Control

numerous methods for control of these robots, yet control-related challenges are to be addressed [92, 93, 94, 95, 96]. Position control is one of the most used methods due to its simplicity. In this approach, the robot tries to follow a predefined motion trajectory [97]. Furthermore, position control is needed to conduct the movements in the first phase along a specific trajectory at a constant speed to improve the ROM capacity of the patient [98, 99, 100, 101]. However, this method cannot be used in the other phases since when the patient applies more than the expected force/torque during rehabilitation, the position method cannot compensate for this disruptive effect, which leads to excessive position tracking errors. Despite this, it should be noted that most of the existing interaction controllers continue to use a position control scheme as an inner control loop, with a corresponding force/torque or motion outer loop applied to complete the interaction controller [102, 103].

To eliminate the problem with position control, impedance control, which provides the desired dynamic interaction between the robot and its environment, is proposed. The physical interaction is expressed as the dynamic relationship between the motion variables of the manipulators and the contact forces that need to be kept within a predetermined safe or acceptable position trajectory while the robot follows the desired motion trajectory [103]. Admittance control uses force as an input and displacement as an output while the robot follows predetermined force. These methods are used in the strength phase to assist/resist the patients [101, 104, 105, 106]. However, these methods require force/torque sensor to measure the force exerted by patients, which

adds extra complexity, i.e., sensor dynamics, cost, and computational load. Moreover, the estimation of impedance parameters peculiar to the individual is another challenge [107].

The dynamic model of robots is usually composed of nonlinear functions of the state variables (joints' positions and velocities), especially in parallel structures. This characteristic of the dynamic model makes the closed-loop control system nonlinear and difficult to solve. The computed torque control (CTC) method requires a good knowledge of the robot dynamics since the dynamic model of the manipulator is used in the loop. Even though, CTC can compensate for non-linearity since dynamic equations are solved in real-time, it creates computational burden [107, 108, 109].

The above-mentioned issues are addressed by using the parameter estimation control method. With this method, the simplification of complex mathematical dynamic equations and modeling of unmodeled noise signals is performed, and a new physical model with functional properties can be obtained with less computation time and acceptable control performance. Furthermore, modeling can be performed online or offline. In the offline method, if the estimated model is not highly accurate, it cannot be able to correctly distinguish between responses caused by known and unknown input signals. In the online method, parameter estimation is done simultaneously within the process, causing a delay in the system response [103, 110, 111]. The fuzzy logic method is also used in the control of rehabilitation robots. Furthermore, it provides good performance for the control of the nonlinear system. Fuzzy logic is similar to human thought and is more systematic than traditional logic systems. Basically, it tries to capture the approximate, imprecise nature of the real world. It has three steps: fuzzification, linguistic rules, and defuzzification. In the fuzzification step, input signals are converted into fuzzy sets with some degree of membership (range from 0 to 1). The main part of fuzzy logic control is controlling the robot using a set of linguistic control rules associated with binary concepts such as fuzzy

inference and computational inference rules. In the defuzzification, fuzzy truth values are converted into output decision values. However, like parameter estimation, it is not robust against unexpected situations during real-time control since its rules are determined previously [112, 113, 114].

In the above-mentioned methods, the patient passively follows the movements of the robot, which follows the previously specified position or force/torque references [115, 116]. Recent studies in robotic therapy show that continuous passive training therapy does not significantly improve motor function. Active participation of patients is considered to be a major factor contributing to neural plasticity and motor recovery [91, 117]. One of the most commonly adopted assistance strategies, the "Assist-as-Needed (AAN)" paradigm, offers the mode of necessary active assistance to stimulate neuroplasticity. The basic principle in AAN is to provide physical assistance only when needed by the patient. If a patient performs a task flawlessly, the robotic assistance is withdrawn. However, if the patient has difficulty or cannot complete the task, the robot provides as much support as the patient needs to perform the task [118]. In other words, AAN is a strategy of regulating auxiliary forces/torques or task difficulty according to patients' disability level or performance in training tasks. There is strong evidence that active participation induces neural plasticity, and therefore, controllers should intervene minimally to promote participation and recovery. In addition, upper extremity rehabilitation using the AAN paradigm is shown to be the most promising technique for promoting recovery [119].

New technological devices and methods are being developed to increase active patient participation by integrating multi-sensory information and allowing the AAN paradigm. However, the implementation of the AAN paradigm is still challenging since determining the level of assistance according to the patient's progress is not to be addressed sufficiently.

1.5 Design of Existing Rehabilitation Robots

Considering the design and development of robotic devices for lower extremity rehabilitation, there are mainly three types of system, i.e., wearable exoskeleton system, ankle platform and balance rehabilitation platform are proposed [7, 87, 120]. Several examples of exoskeleton devices aim to increase the capacity of the lower limbs or reducing user effort [121, 122, 123, 124]. Systems such as AKROD [125], BioMot project [126], KNEXO [127], Lokomat [128], LOPES [129], MIRAD project [130], and REX [131] are used to support individuals with muscle weakness in ADL. However, since the mechanisms underlying human movements and how the designed devices should interact with humans are not fully understood, there is no device that can effectively improve the user's performance. The mentioned systems have difficulties in use because they have a rigid, bulky structure and uncomfortable interfaces, restrict biological joints, and are misaligned with natural joints. In addition, if the exoskeleton does not have enough DoF to work in harmony with human joints, it exerts a residual force on the human limb due to axial misalignment, and this may cause long-term injuries, as well as discomfort [132, 133].

Platform-based robots are grounded and have movable end effector as a rehabilitation platform with one or more DoF. These types of systems employed for the ankle joint focus only on improving the ROM of the joint rather than improving the balance of the patient [87, 134]. Most of the platforms for the ankle joint are in parallel structure, which provides sufficiently high torque for plantar flexion/dorsiflexion, inversion/eversion and adduction/abduction movements of the ankle [135, 136]. There are also systems designed to be serially connected to each other with motor-operated joints [137]. Although serial manipulators are easier to model, using a parallel manipulator in such an application provides advantages in terms of achieving high load-carrying capability, better dynamic performance, and precise positioning. Rutgers ankle is a Stewart platform-type haptic interface that provides 6 DoF resistance

forces to the patient's foot in response to VR-based exercises [138]. Various clinical studies are conducted with this device, showing improvement in patient strength and endurance measurements [139, 140, 141, 142]. In RePAiR, a 1-DoF robotic platform, allows dorsiflexion and plantar flexion movements to patients after stroke. It is demonstrated that the device provides benefits in increasing the muscle strength of the patients, improving the motor control, sensory-motor coordination of the patients, and accordingly, the walking patterns [71]. Only a few of the manipulators developed for ankle rehabilitation are commercialized [101, 143, 144]. These platforms are widely used to strengthen ankle joint movements and improve ankle proprioception. The end effector of the above-mentioned manipulators are only one foot large and have a low weight-bearing capacity; therefore, they cannot be used for balance rehabilitation after ankle treatment has been completed. In addition, the end effector of the proposed systems is not endowed with a sensor; therefore, pressure change measurement on the sole and sensory input under the foot cannot be performed during the ROM rehabilitation.

Posturography, measurement of CoM and balance variables, are tested through static or dynamic techniques for balance evaluation and rehabilitation [145, 146]. Training with static balance platforms is said to be helpful in controlling pressure distribution in patients over time. Static posturography is usually done by using Wii fit [147] board or commercially available force platforms [148, 149, 150, 151]. Studies that evaluated the Wii exercise experience of the patients and their physiotherapists stated that Wii exercise is an amusing and challenging way to improve balance impairment since VR provides continuous feedback [145, 152]. Another study was conducted for 6 weeks with commercially available force platform and VR environment, and the result shows that the CoM points deviation decreased. However, compared to static situations, rehabilitation under dynamic conditions contributes more to improving balance disorders and motor skills [145].

Dynamic platforms require instantaneous dynamic movements, forcing patients to adjust their balance during perturbation [145]. A study shows that dynamic strength platforms are more helpful in restoring postural stability than conventional therapy [137]. Wooden balance platforms are one of the simplest examples, but they do not provide any quantitative measurement [153]. The Bobo Balance platform provides force measurement on top of wooden platforms. However, it is difficult to use for people with severe loss of balance since there is no external support environment or mechanisms for patients while standing [154]. Hunovo [155], Huno [156], Bobo Balance [154], gePRO [157], SMART Balance Master [158], BackinAction [159], Multitest Equilibre [160], Balanceback [161], Tymo [162], and Huber 360 [163] platforms have 2-DoF (roll and pitch) for balance rehabilitation. These systems deliberately put patients in an unbalanced state while patients following the VR game, thus assessing their balance status based on CoM position. During the dynamic rehabilitation process, the angle of the platform can be controlled, and patients are requested to maintain their CoM and posture. Even though ROM can be covered fully by gradually changing the angular position. These systems cannot provide AAN paradigm (even if a patient’s performance improves, the system’s level of assistance remains unchanged) [94, 164]. Furthermore, weight transfer in balance training cannot be provided during PBT due to the lack of sensory input to the foot with these devices. Additionally, the VR environment is the only sensory input proposed with these devices; however, it has been stated in the literature that multi-sensory input is more powerful for mimicking and enhancing ADL [67].

Investigation of APAs and CPAs, which are necessary for proactive and reactive balance control, is claimed to disclose essential aspects of postural control and history of falls. The studies show severe APAs and CPAs deficiencies after evaluating the patient’s condition and muscular activity. However, retraining APAs and CPAs is possible through PBT programs. Despite the benefits of these studies, to the best of

our knowledge, there is no study in the literature that assesses reaction time, CoM, CoP, and the ability to control APAs and CPAs on ankle, balance, and stepping rehabilitation simultaneously. On top of that, the effects of multi-sensory inputs and cognitive control strategies suitable for the use of patients with different severity levels have not been evaluated and utilized. The aforementioned problems and solution has been address in this thesis with I-BaR framework and BalanSENS robotic platform.



CHAPTER II

INTEGRATED BALANCE REHABILITATION FRAMEWORK

The ability of balance is a multivariate concept that integrates both motor and sensory components. Postural control during standing and walking requires multi-modal sensory feedback, e.g., visual, vestibular, and proprioceptive feedback [9]. Sensory inputs have contributions depending on the environment and the motor task performed by the individual, and patients with neurological damage have difficulties in weighing and utilizing sensory inputs [18, 165].

Rehabilitation of balance should be performed with the integration of ankle-foot, balance, and stepping phases in order to enhance activity-based neuroplasticity. Therefore, this thesis proposes an I-BaR framework that adopts the AAN paradigm for balance analysis, rehabilitation and assistance with multi-modal feedforward and feedback signals.

It is known that the sensory input training can positively affect motor control during balance rehabilitation [80, 166, 167, 168]. Multi-modal information provides certain advantages in terms of effective and realistic training to mimic ADL. For instance, the ability of visualization shows spatial aspects, while audio/tactile feedback allows temporal aspects. Studies on the neurophysiology of somatosensory information processing emphasize that multiple cortical and subcortical brain (CNS and PNS) structures take an active role in sensory discrimination tests [67, 68, 169]. Moreover, a stimulator should be placed on the skin of the patient so that muscles with low activity in sEMG measurement can be triggered. VR environment should be employed to improve the effectiveness of the training and patient engagement while serving as

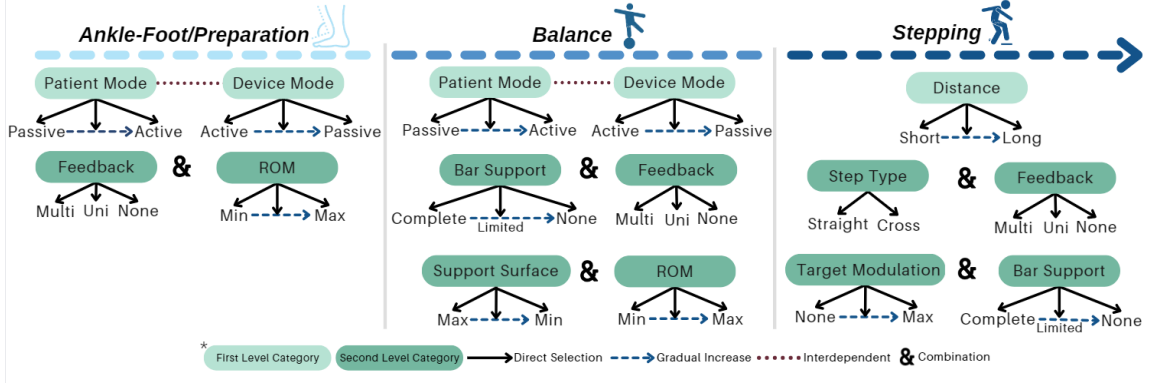


Figure 1: Protocol of I-BaR: (A) Ankle-Foot/Preparation, (B) Balance, and (C) Stepping Phases

visual and auditory feedback. VR tasks should be designed similarly to the activities in which the patient has difficulty, and their difficulty levels should be adjustable.

The I-BaR framework proposes a personalized approach that allows patient-specific difficulty levels in three main phases of rehabilitation (see Figure 1). In the ankle-foot phase, after the mode of the patient see Figure 1A is determined and accordingly, the selected robotic device should be controlled with the AAN paradigm, and feedback mode can be selected according to the patient's needs. Similarly, for the balance phase (see Figure 1B), personalized rehabilitation can be offered according to the patient's capability by selecting different combinations from the bar support and feedback mode Section. Moreover, after the distance is determined in stepping rehabilitation (see Figure 1C), rehabilitation can be performed with the feedback mode and step type Sections mode combinations. In this way, rehabilitation can be provided to the patient at different severity levels in the areas they completely lack. Since each option will be increased gradually throughout the process, he/she can continue their treatment without difficulty.

In this context, the requirements to be considered in order to implement the I-BaR framework and to ensure the development of the patients can be divided into two main groups.

Physical system requirements (Limit reference for ROM:[170], PBT:[171] and Stepping:[172]);

- At least 3-DoF (roll (rotation in x), pitch (rotation in y), and (elevation in height (translational in z)) to mimic ADL
- System should be able to determine applied force/torque based on an individual's parameters (Patient's ROM, ankle torque) to keep the treatment at the limit that the patient can have difficulty continuously
- AAN paradigm implementation on control structure to encourage active participation of the patient
- The system's end effector should include at least four force and haptic sensors on each foot (should cover at least the toes, metatarsals, middle foot, and heel) to quantitatively measure applied force and give haptic feedback accordingly
- Haptic bar should be installed on the system to provide physical assistance on balance and stepping rehabilitation
- Haptic bar should include force and haptic sensors to measure upper extremity support, and this support should be reduced by haptic feedback (if the patient depends to the upper extremity to walk or control the posture)
- Target point on the stepping rehabilitation should be automatically adjustable to provide perturbation/moving target in the x-y plane
- Target point on the stepping rehabilitation should include force and haptic sensor to quantitatively measure walking parameter (GRF on stepping) and increase sole sensation

Rehabilitation requirements;

Table 2: I-BaR Requirements

Exercise method	Limit	Personalization
ROM	20° dorsiflexion, 50° plantarflexion, 10° adduction, 5° abduction, 20° eversion and 35° inversion	Each patient maximum ROM limit should be determined accordingly to torque/force applied on the system (based on the sEMG and force sensor measurement)
PBT	The displacement of CoM and perturbation velocity should be 2-3 cm 0.11-0.18 m/s, respectively	Each patient perturbation speed should be arranged accordingly to the patient postural control level (based on the sEMG, force sensor measurement, CoM deviation and haptic bar/upper extremity support)
Stepping target perturbation	Healthy adult walking speed 1.2–1.3 m/s	Target position should be determined accordingly the second phase response (based on the sEMG, force sensor measurement, CoM deviation and haptic bar/upper extremity support)

- Muscle which has low activity should be triggered with haptic feedback to initiate the motion during all phases
- Each patient should perform the exercises at their specific ankle ROM and torque/force to maintain the patient's safety
- Patient's balance in the mediolateral and anteroposterior directions should be disrupted by large and sudden perturbation to improve plasticity and balance reactions (neuromuscular responses)
- According to the patient's at sub-threshold vibration sensing level, haptic feedback should be provided to improve decreased sole sensation
- Multi-modal feedback (Haptic, audio, visual, physical (anteroposterior and mediolateral) perturbation) should be used to improve sensory-integrating weighting skills

Table 3: Feedback/Feedforward Requirement

Sensor/Method	Limit
Haptic Sensor	Threshold value should be determined according to individual EMG data and weight transfer capacity
Force Sensor	Weight capacity should be at least 100 kg to cover so that 95% of the user population would be compensated.
Position Control	Should be used to prepare the sensorimotor system for motor function when there is minimal or no voluntary motor/muscle function in ankle-foot.
Impedance Control	Should be used to adjust patient's spasticity level (the ability of force application)
AAN Paradigm	Should be implemented in the control structure to provide physical assistance only when needed by the patient and thus to stimulate neuroplasticity
Feedback type	Multi-modal sensory feedback (Audio, haptic visual, vestibular, and proprioceptive) should be implemented on the rehabilitation procedure to integrates both motor and sensory components as whole and thus improve ADL.

- Ankle-foot, balance, and stepping rehabilitation should be performed as a whole, starting from the necessary phase according to the patient's initial physical condition, to achieve gradual independence.

2.1 Ankle-Foot/Preparation Phase

The first phase is ankle-foot/preparation rehabilitation to improve lower extremity muscles while the patients are sitting due to their ankle instability and/or low muscle activation. It aims to prepare the sensorimotor system for motor function, which is essential when there is minimal or no voluntary motor/muscle function in the ankle-foot. The selected robotic platform should be kept in ankle ROM with 0-20° dorsiflexion, 0-50° plantarflexion, 0-10° adduction, 0-5° abduction, 0-20° eversion, and 0-35° inversion limits [170]. The system must ensure that each patient can perform exercises at their ankle ROM limits. Upper limits must be determined according to the impairment levels of the patient, and they should be gradually increased based on their

performance. Such limits can be identified with the use of a robotic platform with a predetermined force/torque based on an individual's parameters, i.e., foot mass, size, and inertial parameters. This torque should have the maximum magnitude to be able to move the individual's foot. The patient can exert reaction force/torque due to spasticity (involuntary muscle contraction) and RoM limits during robotic platform-induced motion. Thus, once the platform cannot move the foot any further, this angle is considered as the patient's spasticity level. Up until this time, rehabilitation is performed with the patient passive mode; however, as the patient progresses in the therapy, his/her contribution to the motion would also be requested at this phase (patient active mode). For instance, moving the ankle to 5° of dorsiflexion with patient's active contribution while the platform supports, and this support can be regulated according to his/her progress.

Moreover, 43% of patients have decreased sensation of their soles after stroke, and it is also being indicated that providing sensory feedback to the sole of the foot (according to their vibration sensing level at sub-threshold) drastically lowers postural sway during standing in older adults with diabetes, and stroke patients [47]. Therefore, a force and haptic feedback sensor should be placed in at least four regions, namely toes, metatarsals, middle foot, and heel, as they are more intense regions [173]. With the force sensor, the pressure applied by the patient to the each region on the sole can be measured, and if the pressure distribution exceeds the able-bodied pressure distribution value in one of these regions, he/she can be stimulated in that region via haptic feedback based on the individually identified sensing threshold. Accordingly, the patient can be aware of the amount of weight-bearing and sole pressure; thus, they can be trained to transfer their weight correctly for fall prevention. With this aim, a personalized preparation therapy program is proposed as an essential stage for I-BaR, which provides multi-modal feedback/feedforward signal and determines the somatosensory level of the patient for the therapist to plan and decide on the

parameters of the next phase.

2.2 Balance Phase

The second phase is balance rehabilitation to improve postural adjustments while the patient is standing on the platform. Although the patient passes this phase after the preparation, they may have difficulty in balancing their posture without physical assistance. Therefore, a haptic bar should be installed on the selected rehabilitation system. This haptic bar should at least include two sensors, force and haptic sensors, to quantitatively measure applied force and give haptic feedback accordingly. In the case of a patient with higher balance loss, the phase should be executed with the full assistance of haptic bars. Throughout the rehabilitation as the patient's postural control improves, an upper extremity support limit value should be established for each patient. When the patient goes over the limit, haptic feedback is applied to warn them to reduce the upper extremity support and help them to control the posture by using the lower extremity rather than relying on the arms.

As the core part of balance rehabilitation, this phase should be utilized to increase sensory and motor integration during task-specific activities. Patient-specific information, such as ROM and proprioception, obtained from the preparation phase determine the required parameters of the therapy at this phase. This rehabilitation can also increase somatosensation during task-specific activity. Within the scope of these interventions, the patient's balance in the mediolateral and anteroposterior directions can be disrupted by giving sudden and fast perturbations with the rehabilitation robot. In PBT, perturbations in different directions and amplitudes are applied to train balance reactions, and various scenarios are used, including simultaneous cognitive and motor activities. This aims to decrease the reaction time and deviation in CoM and improve pressure distribution as well as APAs and CPAs [48, 174]. As in the preparation phase, the AAN control paradigm should be implemented on the

system during PBT so that the patient can engage with the dynamics of the platform independently and actively within the limits of their capacity.

2.3 Stepping Phase

The third phase is stepping rehabilitation to improve walking parameters, i.e., the stance and swing phase's temporal and spatial parameters. While step speed is increasing, improving the stepping accuracy is another target in this phase. It should include assessment and training in stepping to various target distances with robotic platform's help. Step-taking activities are performed to assess the effectiveness of the previous two phases by analyzing the relationship between muscle activity, postural control, movement speed, and accuracy as well as to train walking parameter with Fitts' law. Particularly, at stepping long distances, the patient realizes the movement faster with their own compensation strategies, which reduces the quality of motion, whereas at stepping short distances, their motion is slower, controlled and precise. Current step-taking assessments are based on manual change of the target distance and reaction time measurements (detected by a switch button according to the time between target point and initial position force detection) [40, 42, 175]. Since force measurements and haptic sensors are not implemented at these target points, a kinetic evaluation and improvement in sensory input under the foot cannot be provided. Therefore, in the stepping phase, the distance of the target points should be automatically adjustable within the system to provide perturbation/moving target in x-y plane and these target points should include force and haptic sensors to calculate GRF and increase sensation under the sole, respectively.

2.4 Discussion

Rehabilitation helps with the physiological and functional recovery of motor and sensory skills that have been lost [176]. Recently, it has been stated that investigating the APAs and CPAs, which are required for proactive and reactive balance control,

may reveal critical information about postural control and fall history. Early muscle activation and decreased CoM deviation provided vital evidence that retraining of APAs and CPAs is achievable for stroke and elderly population. The literature provides the fundamentals of the development of PBT to enhance postural control and balance in patients to prevent fall. The therapy procedure used for large body structures, such as the lower extremities and trunk, requires a lot of physical effort for therapists and patients, and the effectiveness of rehabilitation depends on the therapist's personal knowledge and experience [86]. Robotic rehabilitation studies are increased significantly to overcome these difficulties since they are recognized as efficient and provide precise assessment of kinematic and dynamic parameters as well as objective evaluations that enhance the treatments. Yet, there are still challenges to be addressed, such as limited accessibility of present robotic devices due to cost, the training of all patients at the same level of difficulty, and insufficient personalized approach [83, 177].

The aforementioned limitations have been addressed in a separate manner. For instance, in ankle rehabilitation, robotic platforms are frequently used for patients with higher severity levels to improve ankle ROM, strength, and proprioceptive. Although these studies are sufficient to improve ankle instability, PBT is still required for these individuals to improve their ADL. Ankle rehabilitation robots cannot be used in balance rehabilitation because they only have the surface area to fit one foot and have a low weight capacity compared to the whole body weight.

Moreover, in balance rehabilitation, treatment is implemented with either a static or dynamic system. Although static balance systems are useful in evaluating the patient's balance level, dynamic anteroposterior and mediolateral perturbations are required to improve plasticity in patients. Furthermore, in the dynamic balance systems, due to the lack of sensors on the end effector, these systems cannot train patients to control their weight transfer to avoid falls. In particular, the pressure

change in certain regions of the foot (at least four major parts heel, middle foot, metatarsals, and toes) cannot be measured and corresponding vibrotactile feedback cannot be applied under the foot. Accordingly, such systems are not able to improve decreased sensation on the sole.

As the third main treatment, stepping rehabilitation is implemented to improve walking parameters, stepping accuracy and stepping ability to various target distances. Fitts' law is used to explore the role of motor planning processes. According to the Fitts' law, the target's width and distance are correlated and will determine how long it takes to go quickly to a particular region. Furthermore, this law is implemented on the CSRT test to analyze the time to complete the foot-reaching task and improve the walking parameters. Various target points are placed on the environment, and the subject must step on the target accordingly to the given feedback. These target points just include a switch to determine reaction time. In other words, they are unable to assess GRF for haptic stimuli and train the patient to control pressure distribution.

It is essential to associate the aforementioned rehabilitation method with multi-modal feedback since it enables the learning of several aspects of a movement simultaneously and mimics daily life, e.g., visual, pressure, audio, vibration, and proprioception. These feedback information not only promotes the development of plasticity but also provides compensation for the loss of motor function caused due to a compromised neuromuscular system. However, to the best of our knowledge, there is no study to prove the importance of multi-modal feedback in the improvement of postural adjustments and reaction time.

The three-phase framework, I-BaR, is proposed to address the aforementioned drawbacks of current rehabilitation approaches. First of all, all these three separate treatments should be implemented as a whole in a single system to improve foot/ankle functionality and postural adjustments (APAs and CPAs) for achieving high-quality

walking based on objective assessment. Moreover, in all these phases, using the multi-modal feedback for improving sensory weighting skills, the AAN paradigm for modulating the level of assistance according to patients' progress and mimicking the situations that the patient encounters in ADL should be implemented as a whole for an effective assessment and rehabilitation to achieve gradual independence eventually. Therefore, the I-BaR framework offers an effective solution as a whole with these properties to achieve personalized balance rehabilitation for different disability levels.

In this context, the requirements to be considered in order to implement the I-BaR framework and to ensure the development of the patients can be divided into two main groups and they are summarized in Table 2-3.

In summary, this chapter explains the necessity of the proposed I-BaR framework, which includes:

- Ankle-foot Rehabilitation: ankle-foot muscle activation, sole, joint, and movement sensations are developed while sitting,
- Balance Rehabilitation: sensory weighting skills are developed from motor learning by using multi-sensory input during PBT and help gradual independent,
- Stepping Rehabilitation: walking parameters are improved during step-taking activities to target points with adjustable distance.

CHAPTER III

MECHATRONIC DESIGN OF THE BALANSENS SYSTEM

The human-centered design principle emphasizes the ergonomic approach to design: If an object, device, or ecosystem is designed for human use, the designer must consider the physical and mental capabilities of the user. The aim is to provide an ideal harmony between the designed product (object, system, or environment) and its users within the scope of the working environment. Critical elements of ergonomic design should be determined on a case-by-case basis. Nevertheless, some initial criteria can be determined, such as functional efficiency, convenience, comfort, safety, and working conditions. Every person is unique (in terms of size, morphology, and all other physical characteristics), and human-centered design requires an understanding of this diversity. As a result, the system must be developed to be comfortable for every user, adjustable and available in various sizes. Requirements of each individuals are considered in the design process of the system, as given in Table 4. The lower and upper limits are determined with 5th percentile for female proportions and the 95th percentile for male proportions, respectively, so that 95% of the user population would be compensated [178].

With this aim, this thesis proposes a completely personalized rehabilitation system for patients with different severity levels or disorders. It integrates three main phases of I-BaR framework, i.e., ankle-foot, balance, and stepping rehabilitation. There are three different modules on the BalanSENS system (see Figure 2A), which can be customized for different rehabilitation purposes based on the patients' needs. The chair placement and the first step target points are included in the first module (see

Table 4: Design criteria consideration

	Minumum(mm)	Maximum(mm)
Height	1505	1855
Eye height	1405	1745
Sitting eye height	685	845
Knuckle height	660	825
Knee height	455	595
Hand breadth	70	95
Foot length	285	215
Foot breadth	80	110
Buttock–popliteal length	435	550
Body weight (kg)	44	100

Figure 2B-C) that can be used in ankle-foot and stepping rehabilitation, respectively. In the second module, there is a parallel manipulator, which can be used for two tasks. First, it can give perturbation (in ankle-foot and balance rehabilitation), and second, patients can use the manipulator’s end effector as the second stepping region. The third module includes the VR setup, and the third step targets points. With this aim, our research proposes a completely personalized rehabilitation system for neurological patients with different severity levels or disorders. It integrates three main phases of I-BaR, i.e., ankle-foot, balance, and stepping rehabilitation. Each region and customization for the I-BaR protocol is detailed in Sections 3.1, 3.2, and 3.3.

3.1 Ankle Rehabilitation

The first step module is removed during ankle rehabilitation, and a chair is placed in agreement with the ergonomic parameters (see Table 4, Figure 2B-C). Hip-popliteal length, hip width, and shoulder height are some of the important parameters for personalized chair design criteria. The hip-popliteal length influences the depth of the chair; the more the depth deviates from the required length, the more difficult it is for users to acquire support and comfort from their backs. A depth of 300 mm and a height of 500 mm is adequate to support patients’ ischial tubercles and backs,

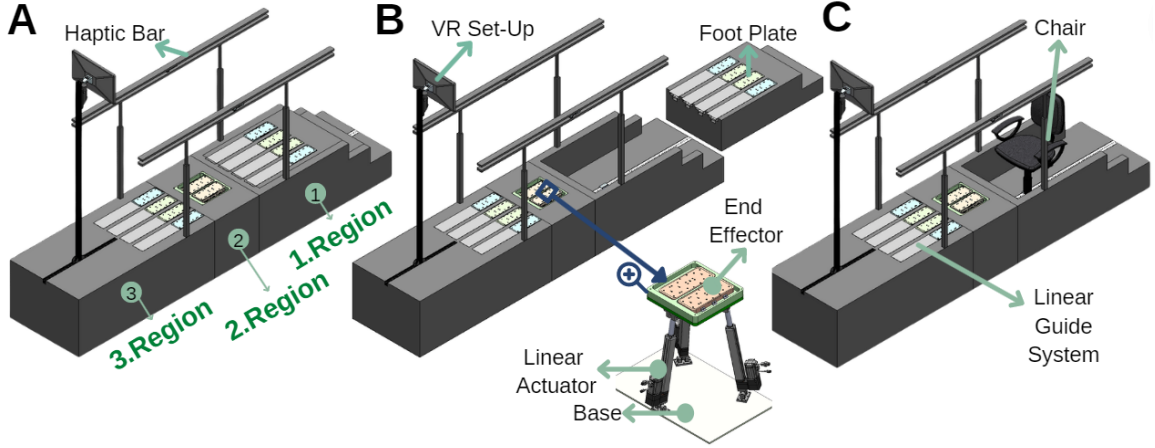


Figure 2: Representation of (A) BalanSENS (B) the removal of the first region (C) chair placement

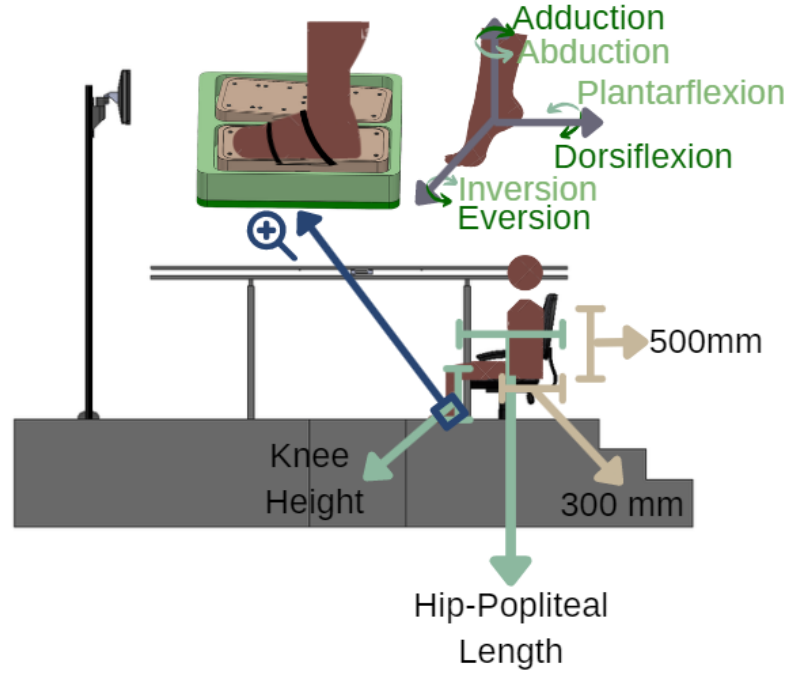


Figure 3: Ankle Rehabilitation Set-up

respectively. Additionally, the distance between the knee, hip-popliteal, and parallel manipulator distance must be adjustable since each patient has a different body size. Similarly, the chair's height should have a range between 435 and 550 mm (see Figure 3) [178].

The 3-DoF parallel manipulator (see Figure 2B) provides a dynamic environment

during ankle and balance rehabilitation. It is equipped with three linear actuators mounted between the end-effector and the base of the robot. It is able to rotate in roll (α) and pitch (β) angles and moves along the z-axis. These DoF are chosen to provide mechanical simplification of the system but still provide a quality balance rehabilitation [169]. Roll and pitch are some of the ankle's most used angle changes, and the system's elevation in the z-axis also covers most of the active activities in real life. The ankle ROMs that can be realized with the system are limited to a healthy individual's ankle. These ROMs (see Figure 3) are limited with 0-20° dorsiflexion, 0-50° plantarflexion, 0-10° adduction, 0-5° abduction, 0-20° eversion and 0-35° inversion [170]. Upper limits are determined according to the severity levels of the patient and are gradually increased to the maximum limit based on their performance. Thus, ankle ROMs and proprioception are gradually improved, as seen in the I-BaR ROM specification Section.

The end effector of the parallel manipulator (see Figure 4A) is designed as a two-layer structure retaining multiple load cells and vibration motors. The upper platform (see Figure 4B) is responsible for evaluating the pressure obtained from the sole of the foot. There are two separate foot plates (see Figure 4B-E) for the right and left feet on the upper platform. Each foot plate is divided into four regions, i.e., toes, metatarsals, middle foot, and heel. These regions are known as the higher amount of weight bearing on the sole and are used as references when determining the placement of the load cells [173, 179]. Furthermore, it is indicated that 43% of patients decreased sensations of the soles of their feet [47]. For these reasons, as shown in Figure 4E, vibration motors are placed on each of the regions. These motors are activated if the patient's weight transfer/pressure is insufficient during proprioception training. With this feedback, the patient can sense the amount of weight transfer and pressure on their soles and be trained to transfer their weight correctly to prevent falls. This feedback specification and VR environment can be used in ankle rehabilitation as

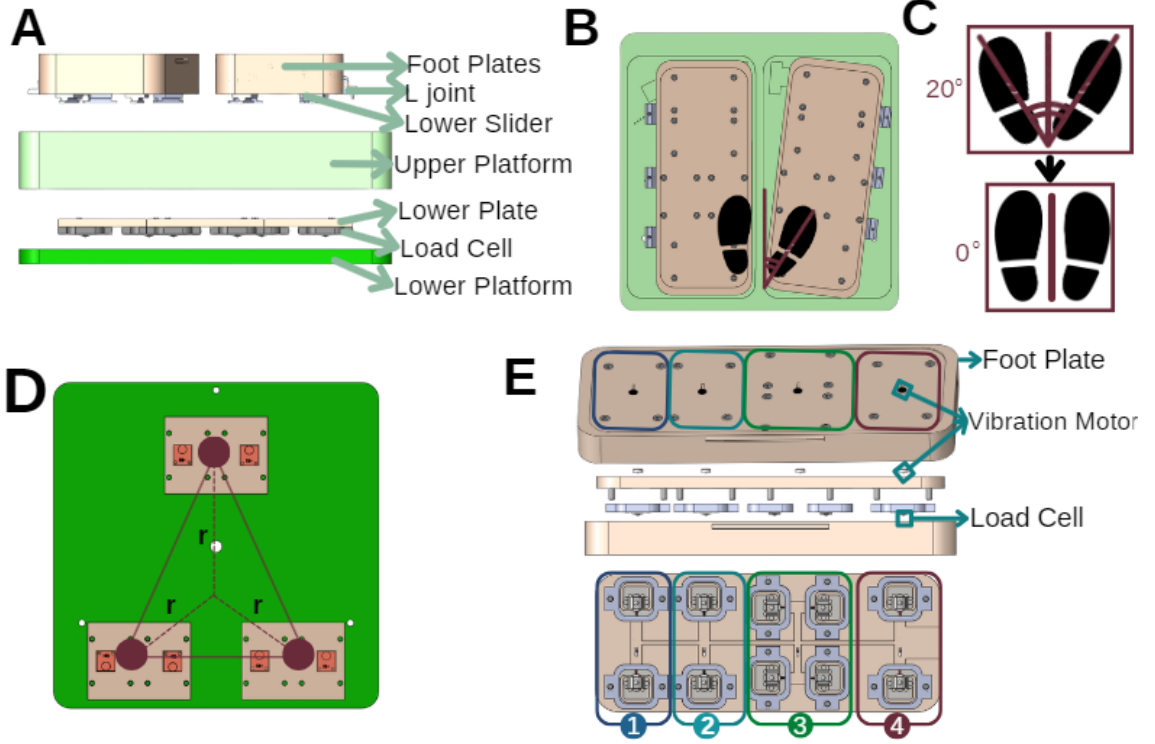


Figure 4: (A) End effector layers, (B) Upper platform, (C) Foot span-support area relationship, (D) Lower platform, and (E) Foot plate

multi, uni, and no feedback options.

3.2 Balance Rehabilitation

The 3-DoF parallel manipulator (see Figure 2B) can measure reaction time, pressure distribution and CoM. Thus, the platform quantitatively assesses patients' balance ability and their cognitive/physical responses to compensate for the impaired balance against the perturbations with various tasks and multiple motor inputs. As in Section 3.1, the platform movement initiates with a limited ROM, and as the level progresses, the angular deviation increases up to the limit specified according to the maximum ankle anthropomorphic. Thus, once patients begin to control the preliminary minor perturbations, they can balance the major perturbations better, i.e., as shown in the I-BaR ROM Section limit.

The foot rotation angle (FRA) on the upper platform's foot plates is adjustable

between $0-20^\circ$ (see Figure 4B-C). As the angle between the FRA decreases, the movement becomes more difficult due to the decrease in the support surface. In balance rehabilitation, the foot rotational angle value gradually reduces, as explained in the support surface specification within I-BaR so that patients can experience gradual recovery. The lower platform (see Figure 4A-D), located just below the upper platform, is responsible for evaluating patients' CoM. This instantaneous CoM measurement is used to provide feedback to the control architecture to determine the patients' balance capacity and adjust the level of assistance.

The required upper extremity support by the patients during PBT is calculated quantitatively with the haptic bar, which contains load cells and vibration motors (see Figure 5). The width of the haptic bar is designed according to the hand width of an average person (for ease of grasp), and their height can be adjusted according to the needs of the patients (600-900 mm) [178]. First of all, patients with high balance loss are asked to balance their CoM during PBT with the complete help of haptic bars. As the patient's postural control improves, an upper extremity support threshold value is determined for each individual and constantly updated during rehabilitation (as seen in the I-BaR haptic bar specification Section). If the patient exceeds the threshold, haptic feedback is applied to reduce the support received and regain their balance. As patients progress in their balance capacity during rehabilitation, they are asked to receive upper limb support only when necessary. Consequently, it is intended that they use their affected lower limb rather than relying mainly on their upper limb, as they are unable to apply excessive force for weight support.

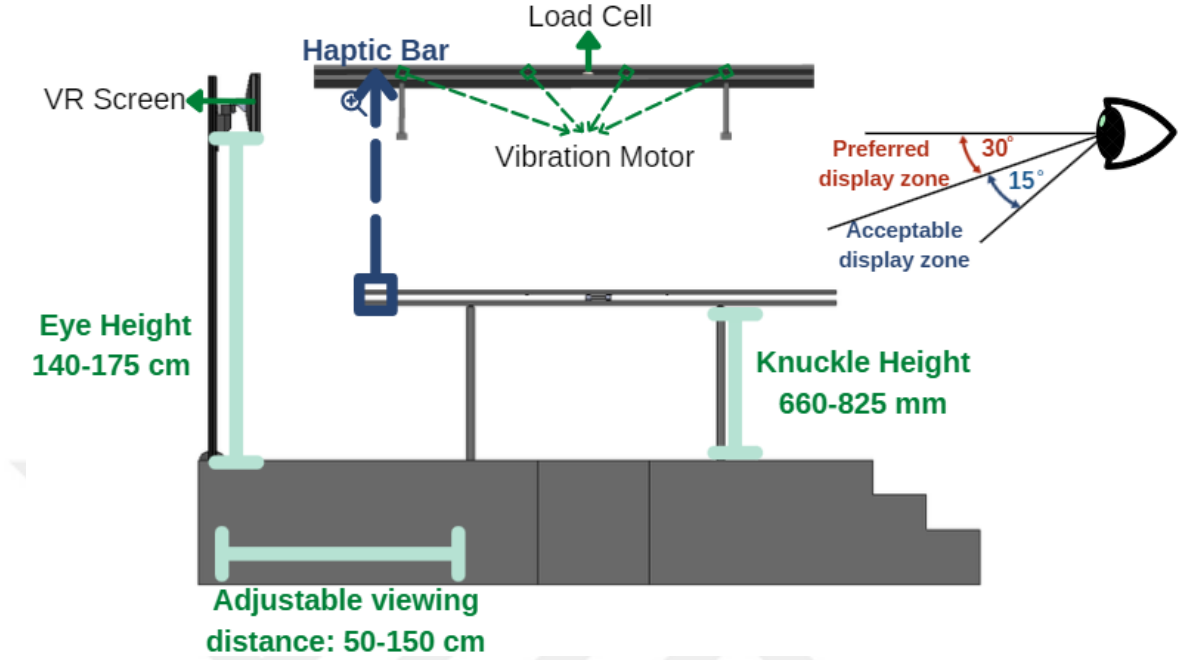


Figure 5: Representation of VR screen and haptic bar adjustment

3.3 *Stepping Rehabilitation*

While walking rehabilitation primarily aims to enhance muscle contribution and activation during continuous walking, stepping rehabilitation focus diverges. Its objective is to enhance the patient's decision-making process rather than directly improving their walking performance. Stepping rehabilitation is used to enhance the ability to step over specific target distances, step accuracy, and gait characteristics. Furthermore, while treadmill-based balance rehabilitation proves beneficial for individuals with higher impairment levels, the focus on stepping rehabilitation caters to a different demographic. Specifically, stepping rehabilitation's target group comprises individuals capable of walking but exhibiting slower reactions and sensory processing.

The modifiable stepping target points are essential since variations in APAs and muscle and strength-duration are influenced by task distance: the more challenging the task, the later postural muscle activity begins. Particularly, at stepping long

distances, patients realize the movement faster with their own compensation strategies, which reduces the quality of motion, whereas at stepping short distances, their motion is slower, controlled, and precise. This compensation can be assessed with BalanSENS by analyzing the relationship between muscle activity (EMG), postural control (CoM deviation control), movement speed, and accuracy (weight shift and pressure distribution). Moreover, previous research on balance control has shown that adopting an external focus of attention compared to an internal focus of attention results in better performance during motor skills [40]. For the reason stated, the external starting signal can be provided visually with VR and auditorily with a sound warning in stepping rehabilitation. The warning selection can be made as multi, uni, and no feedback option (see Figure 1). The distance between the stepping locations can be modified between 15-75 cm (as seen in Figure 6A-B and Figure 1C) better to comprehend the correlation between movement speed and accuracy.

Foot plates are strategically positioned on designated target points, enabling the calculation of ground reaction force (GRF) and providing haptic feedback as an additional external signal in cases where patients are unable to initiate a step following the initial external signal. The muscle activity is determined according to two measurement parameters: the time window between the onset of the muscle activity measured by wireless and mobile EMG service and the through-the-movement kinematics. The methods used in previous research involved manual adjustment of the distance between the target points. The BalanSENS incorporates a linear guide system for adjusting the position of the target points. The utilization of the linear guide system presents an additional benefit by introducing perturbations during the patient's progression toward the target points, facilitating a dynamic rehabilitation approach that closely emulates ADL. The velocity of these perturbations can be customized based on the severity level of the patient, allowing for personalized rehabilitation interventions.

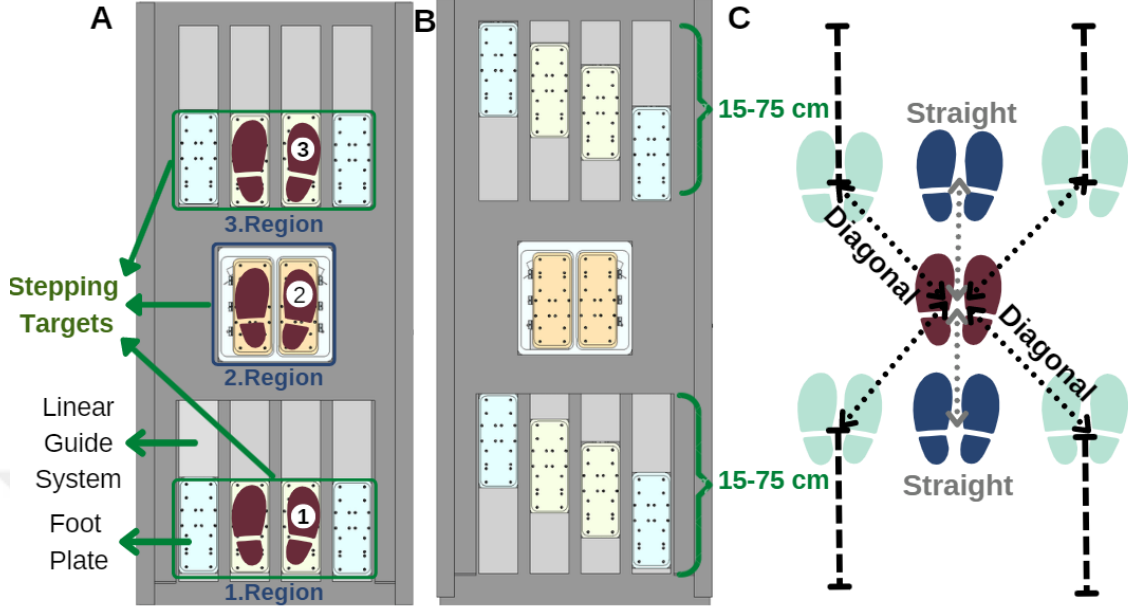


Figure 6: Representation of (A) step rehabilitation set-up (B) stepping distances and (C) stepping measurement process

3.4 Design of Task-Based Virtual Reality Environment

The VR environment of the system is used to increase the effectiveness of the treatment and make the rehabilitation process enjoyable. As shown in Figure 5, the VR environment is placed in front of the patients to ensure proper body-eye coordination. By incorporating VR games into the rehabilitation process, patients can receive comprehensive feedback through visual and auditory stimuli. The tasks within the VR environment are meticulously designed to replicate activities that pose challenges for patients, and their difficulty levels can be precisely customized to cater to individual patient needs. In this way, the patient can respond more consistently to subsequent imbalances in PBT. The reaction time of the patients may be shortened since the control strategies are mostly acquired through learning.

Patients encounter difficulties in effectively transferring their body weight onto the soles of their feet. The VR environment shown in Figure 8A is designed for weight transfer treatment. The pressure on the load cells in the foot areas is evaluated

according to motion specifications. If the value is appropriate, it is shown as green, but the colors turn towards red as the pressure deviates increases. With the second VR environment shown in Figure 8B, patients are asked to reach the targeted points by controlling their CoM. The colors of the dots change according to the gradual pressure change between green and red. Similarly, in Figure 8C, patients are expected to collect the cubes from the table area by controlling their CoM during PBT. In Figure 8D, patients are expected to collect cubes and avoid cylindrical shapes by controlling their right-left foot weight distribution. For each game, the assistance given by the 3-DoF parallel manipulator can be arranged depending on the patient's need.

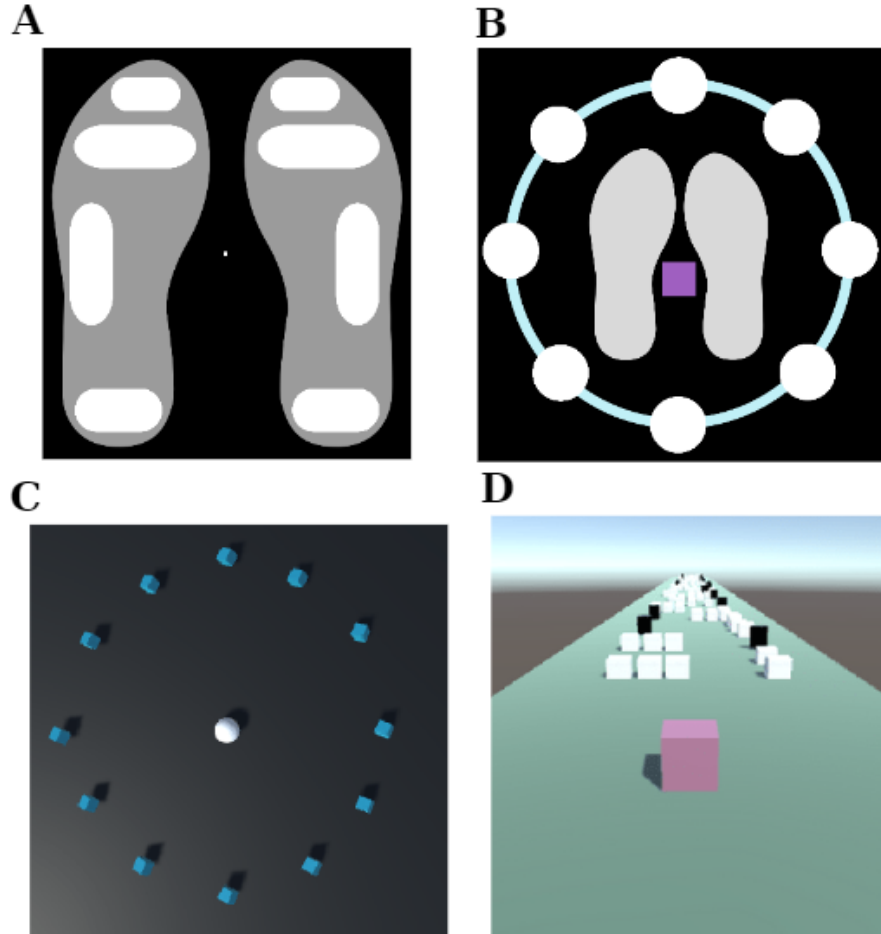


Figure 7: VR games

3.5 *Wearable Sensors*

One of the most beneficial rehabilitation techniques is task-oriented training that provides feedback from various sensory inputs, such as pressure, vibration, and proprioception [9]. These inputs aid in overcoming the loss of motor function caused by the damaged neuromuscular system by developing compensating mechanisms and strategies. One way to address this problem is with VF since it provides haptic stimuli to the points where the loss of sensation and triggers the required region. That's why, VF rings (see Figure 8A-C) are placed on the patient. If the patient does not show muscle activity (measured with EMG signal), VF rings will be used to stimulate/trigger the patient's muscles with a feedforward signal (see Figure 9). In these rings, the vibration motor is placed around the soft tissue and attached to a belt so it is not in direct contact with the patients, and the diameter of the ring can be adjusted. Similarly, the deviation in the trunk, arms, and legs of the patients are measured with the IMU ring (see Figure 8A-B).

3.6 *Discussion*

Robot-assisted rehabilitation delivers more effective training and unbiased evaluation than conventional rehabilitation since it can provide repeated and high-quality task-specific therapy at a relatively low cost and with fewer therapists [82]. Different systems and tests have been developed for different disease types and severity levels. However, since these systems/tests address specific severity levels, they cannot be used in the entire rehabilitation process. Ankle robots have been produced for patients who cannot stand and move their ankles. However, most of them cannot be used in balance rehabilitation as stated in the introduction Section. Similarly, balance robots have been produced to improve postural adjustments and reduce the fall rate for patients who can partially support their balance. Another method used to provide improvement in the development of postural adjustments is CSRT. Here, it

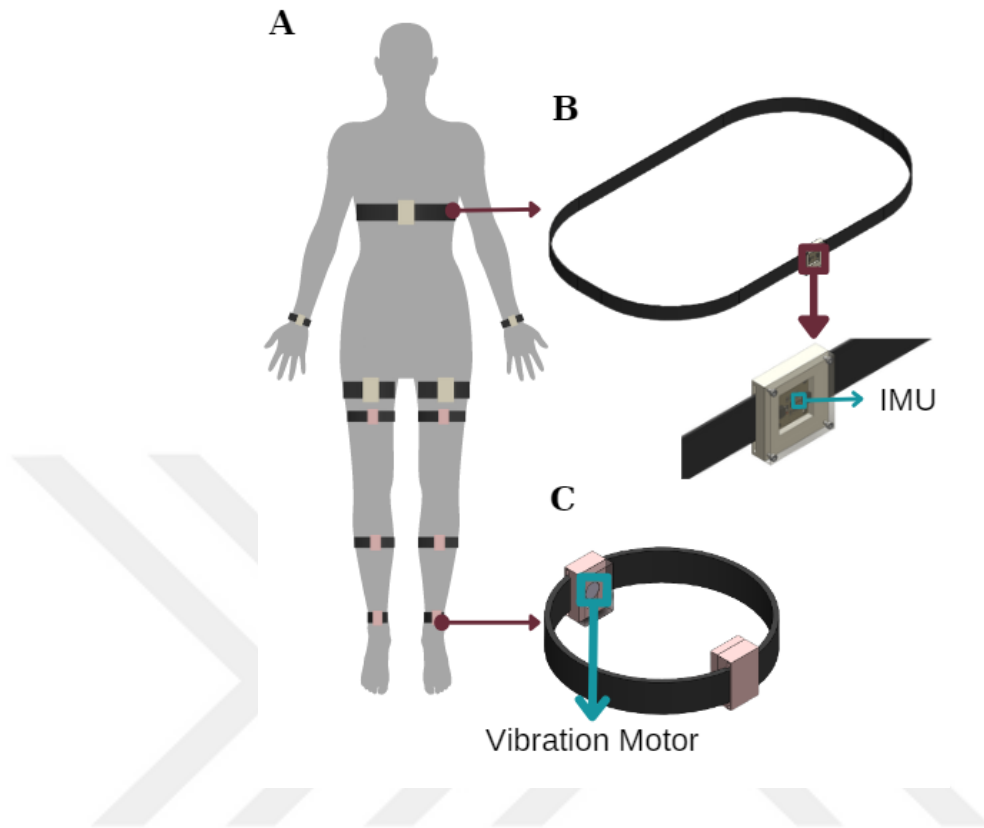


Figure 8: Wearable Sensor Placement

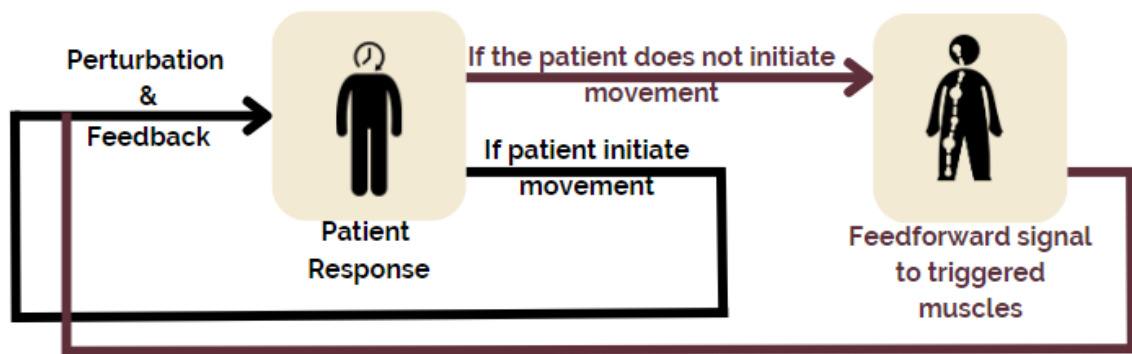


Figure 9: VF ring stimulation process

is aimed that patients who have the capacity to take a step will take steps to certain target points according to the external signal given to analyze postural adjustment parameters. Despite the benefits of robotic therapy, these target points are adjusted manually and do not have sensors to measure GRF. But, both sensation loss on the soles of the feet and decreased muscle activation are observed in patients. These

can be addressed by placing a VF sensor on the end effector of the system and the targeted muscle group during PBT. Moreover, only VR game is used as feedback on these systems/tests thus, they cannot improve re-weighting skills between different senses. This is important due to in real-life humans face multiple stimuli at the same time.

BalanSENS is capable of providing multi-modal sensory input. It aims to test these feedback and feedforward signals for each patient specifically and advance from single-mode to multi-modal applications for neuromotor rehabilitation. The first one is the vibration motors placed on the footplates as mentioned in Section 3.1; this sensory feedback will provide a perturbation by applying VF under the sole. Secondly, it is the vibration motors added to the haptic bar as mentioned in Section 3.2 with these motors; it is aimed to focus on the lower extremity muscles by minimizing the upper extremity support of the patient in the balance treatment. Thirdly, with VR games, the patient will receive visual and auditory feedback. This helps in ADL since, in real life, the majority of feedback comes from either visual or auditory. On top of that, VF rings are placed on the patient to stimulate the muscle with a feedforward signal in accordance with the given perturbation. Lastly, with the parallel platform and linear guide system that can provide anteroposterior and mediolateral perturbations, patients will be trained in rehabilitation according to the dynamic environmental conditions they encounter in ADL.

The proposed BalanSENS system can encourage the active participation paradigm from the CNS to the PNS by integrating the aforementioned multi-sensory information with the co-improvement of sensory and motor functions. In order to meet the requirements specified in the I-BaR framework, three main regions have been created on the system that can be changed. The I-BaR framework and BalanSENS system offer a protocol for individuals with different disease levels to participate in

the rehabilitation process. It is explained in three phases, namely, ankle-foot, balance, and stepping. The first phase is ankle-foot, as patients with severe conditions cannot stand and their ankle muscles are weak, which aims to increase mobility, joint proprioception, and sole sensation. As the patient's ankle condition improves, the balance phase is started. This phase aims to enhance sensory weighting abilities and provides gradual independence via PBT. The stepping phase aims to enhance walking characteristics by performing stepping in target points with distance modulation.



CHAPTER IV

SIMULATION EVALUATION OF BALANSENS SYSTEM

4.1 *Kinematic and Dynamic Analysis*

This Section defines the kinematic analysis of the 3 DoF parallel manipulator constructing the vector loop equations and the system's dynamic analysis based on the Euler-Lagrange method.

The architecture of the 3-DoF parallel manipulator is shown in Figure 2. It comprises an end effector (a two-fold layer, e.g., upper and lower platform), a fixed base, and three supporting limbs/linear actuators with an exact kinematic structure. Figure 10 presents the schematic representation of the 3 DoF parallel manipulator where the bold triangular lines describe the base and end effector.

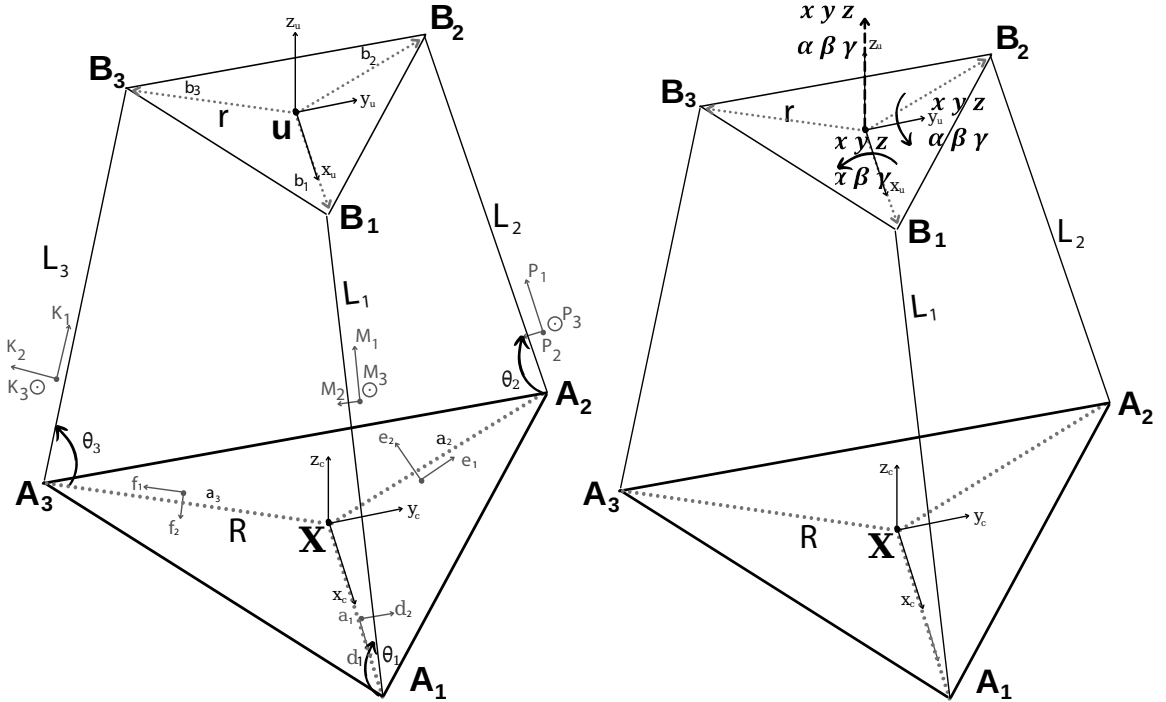


Figure 10: Schematic representation of 3 DoF parallel manipulator

The base is attached to the end effector by three identical PRS linkages, i.e., a sequence P (Prismatic) joint, an R (Revolute) joint, and an S (Spherical) joint, where a linear actuator actuates the P joint and others are passive joints. The end effector changes its orientation on 6-axis, but only 3 of them are controllable, and others are solved with the constraint equation, which is explained in Section 4.1.1. Two DoF of the manipulator, regulated by roll (α)-pitch (β) angles, are assigned for the orientations in two directions, while one DoF of the manipulator is allocated for the translation in the z-direction.

The mobility criterion of the proposed 3-DoF parallel manipulator is proven with Eq. (1) [180].

$$DoF = 6(n - g - 1) + \sum_{i=1}^g f_i \quad (1)$$

Eq. 1 calculates the DoF of the system. ' n ', ' g ', and ' f_i ' represent each joint's number of bodies, joints, and DoF, respectively. In the proposed platform, ' n ' is equal to 8 (3 linear actuators, three passive joints, one end effector (upper and lower platforms are mounted each other), and fixed based), ' g ' is equal to 9 (3 ball and socket, three prismatic, and three revolute), ' f_i ' for ball and socket joint is 3, whereas for prismatic and revolute joints is equal to 1. Eq. 1 results in 3, Eq. 2, which also proves that the proposed robotic platform has 3-DoF.

$$DoF = 6(8 - 9 - 1) + (3(1) + 3(1) + 3(3)) = 3 \quad (2)$$

4.1.1 Kinematic Analysis of the Manipulator

The proposed solution for the kinematic analysis utilizes three vector loop equations to construct the geometric relations among the frames. As depicted in Figure 10, the end-effector and the base frame are geometrically expressed as a means of two equilateral triangles connected by frames of three linear actuators. Here, the symbol r represents the distance between the corners of the triangles and the center of the platform, also symbol θ_i defines the positioning of the actuators on the base frame.

Symbols X and u are the centers of the base platform and upper platform (end-effector) as depicted in Figure 10, respectively. A fixed coordinate frame ($\mathbf{X}$) is established on the base platform and the coordinates of the frame are named x_c , y_c , and z_c .

Coordinates A_1 , A_2 , and A_3 of the corners of the base frame are interpreted with respect to $\mathbf{X}$ frame by means of the position vectors in Eq (3).

$$A_1 = \begin{bmatrix} R \\ 0 \\ 0 \end{bmatrix}, A_2 = \begin{bmatrix} -1/2R \\ \sqrt{3}/2R \\ 0 \end{bmatrix}, A_3 = \begin{bmatrix} -1/2R \\ -\sqrt{3}/2R \\ 0 \end{bmatrix} \quad (3)$$

The coordinates of the corners of upper platform depicted in Figure 10, b_1 , b_2 , and b_3 with respect to $\mathbf{u}$ frame are described in Eq (4).

$$b_1 = \begin{bmatrix} r \\ 0 \\ 0 \end{bmatrix}, b_2 = \begin{bmatrix} -1/2r \\ \sqrt{3}/2r \\ 0 \end{bmatrix}, b_3 = \begin{bmatrix} -1/2r \\ -\sqrt{3}/2r \\ 0 \end{bmatrix} \quad (4)$$

The frame $\mathbf{u}$ can be expressed with respect to the $\mathbf{X}$ frame by using Euler Angle Representation. In Matrix (5), symbols (α) , (β) and (γ) represent roll, pitch, and yaw angles, respectively. Accordingly, the coordinates, b_1 , b_2 , and b_3 stated with respect to the $\mathbf{X}$ frame by means of the Euler Angle transformation, T is expressed in matrix notation as:

$$T = \begin{bmatrix} C_\alpha C_\beta & C_\alpha S_\beta S_\gamma - S_\alpha C_\gamma & C_\alpha S_\beta C_\gamma + S_\alpha S_\gamma & X_U \\ S_\alpha C_\beta & S_\alpha S_\beta S_\gamma + C_\alpha C_\gamma & S_\alpha S_\beta C_\gamma - C_\alpha S_\gamma & Y_U \\ -S_\beta & C_\beta S_\gamma & C_\beta C_\gamma & Z_U \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (5)$$

The position vectors b_1 , b_2 , and b_3 with respect to fixed frame ($\mathbf{X}$) are typified with symbols B_1 , B_2 , and B_3 , respectively. Here, C and S symbolize the cosine and

sine functions, respectively.

$$\begin{bmatrix} B_i \\ 1 \end{bmatrix} = \begin{bmatrix} T \end{bmatrix} \begin{bmatrix} b_i \\ 1 \end{bmatrix} \quad (6)$$

$$B_1 = \begin{bmatrix} rC_\alpha C_\beta + X_U \\ rS_\alpha C_\beta + Y_U \\ -rS_\beta + Z_U \end{bmatrix} \quad (7)$$

$$B_2 = \begin{bmatrix} \frac{-rC_\alpha C_\beta + \sqrt{3}r(C_\alpha S_\beta S_\gamma - S_\alpha C_\gamma)}{2} + X_U \\ \frac{-rS_\alpha C_\beta + \sqrt{3}r(S_\alpha S_\beta S_\gamma + C_\alpha C_\gamma)}{2} + Y_U \\ \frac{rS_\beta + \sqrt{3}rC_\alpha S_\gamma}{2} + Z_U \end{bmatrix} \quad (8)$$

$$B_3 = \begin{bmatrix} \frac{-rC_\alpha C_\beta - \sqrt{3}r(C_\alpha S_\beta S_\gamma - S_\alpha C_\gamma)}{2} + X_U \\ \frac{-rS_\alpha C_\beta - \sqrt{3}r(S_\alpha S_\beta S_\gamma + C_\alpha C_\gamma)}{2} + Y_U \\ \frac{rS_\beta - \sqrt{3}rC_\alpha S_\gamma}{2} + Z_U \end{bmatrix} \quad (9)$$

Actuators' positions on $\mathbf{X}$ frame identify the constraints of the kinematics as illustrated in Fig 10A. The constraint equations are stated for each actuator's static body as $y_c = 0$ (first actuator's constrain plane), $y_c = \sqrt{3} x_c$ (second actuator's constrain plane) and $y_c = -\sqrt{3} x_c$ (third actuator's constrain plane). The constraint equations are derived as in Eqs (10-12).

$$rS_\alpha C_\beta + Y_U = 0 \quad (10)$$

$$\frac{\sqrt{3}r(S_\alpha S_\beta S_\gamma + C_\alpha C_\gamma) - rS_\alpha C_\beta + 2Y_U}{-2\sqrt{3}} = \frac{\sqrt{3}r(C_\alpha S_\beta S_\gamma - S_\alpha C_\gamma) - rC_\alpha C_\beta + 2X_U}{2} \quad (11)$$

$$\frac{2Y_U - rS_\alpha C_\beta - \sqrt{3}r(S_\alpha S_\beta S_\gamma + C_\alpha C_\gamma)}{2\sqrt{3}} = \frac{2X_U - rC_\alpha C_\beta - \sqrt{3}r(C_\alpha S_\beta S_\gamma - S_\alpha C_\gamma)}{2} \quad (12)$$

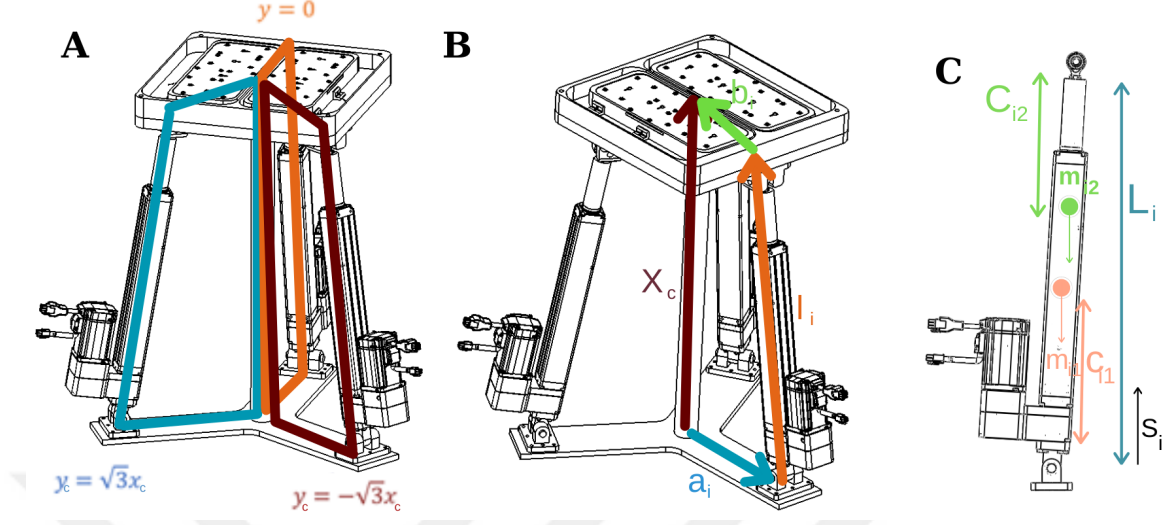


Figure 11: (A) Geometric constraints (B) Vector loops for the kinematic and dynamic analyses (C) Linear actuator link variables

Eq (13) and Eq (14) are derived to represent the constraint equations in $\mathbf{u}$ frame by adding and subtracting Eq (11) and Eq (12).

$$S_\alpha C_\beta = C_\alpha S_\beta S_\gamma - S_\alpha C_\gamma \quad (13)$$

$$X_U = \frac{r(C_\alpha C_\beta - S_\alpha S_\beta S_\gamma - C_\alpha C_\gamma)}{2} \quad (14)$$

Eq (13) satisfies the equality for any integer of n ; however considering the physical limitation of the system requirement, the summation of α and γ cannot exceed zero as expressed in Eq (15).

The constraint Eqs (15-17) are rearranged to diminish the number of unknowns from five (x , y , $\text{roll}(\alpha)$, $\text{pitch}(\beta)$ and $\text{yaw}(\gamma)$) to two ($\text{roll}(\alpha)$ and $\text{pitch}(\beta)$). The relations among the x , y , z , $\text{roll}(\alpha)$, $\text{pitch}(\beta)$ and $\text{yaw}(\gamma)$ angles expressed in Eqs (13-14) are utilized to determine the constraint Eqs (15-17) to characterize the upper platform by X_U and Y_U depicted in Figure 10.

$$\alpha + \gamma = n\pi, \quad \text{where } n = 0 \quad (15)$$

$$X_U = \frac{-r(1 - C_\beta)C_{2\alpha}}{2} \quad (16)$$

$$Y_U = \frac{r(1 - C_\beta)S_{2\alpha}}{2} \quad (17)$$

The connection between the upper and lower platforms is carried out via three linear actuators. Considering the kinematic point of view, the vector-based representation of the actuators' positions in the space is defined by their lengths, l_i , where $i = 1, 2, 3$. The Pythagorean theorem allows us to establish the fundamental relation in Euclidean geometry among the three sides of a right triangle as depicted in Fig 10B and the formula to express the length of each actuator is presented in Eq (18).

$$l_i = \sqrt{(B_{ix} - A_{ix})^2 + (B_{iy} - A_{iy})^2 + (B_{iz} - A_{iz})^2} \quad (18)$$

Accordingly, the length of each actuator is expressed as a function of three independent variables of roll(α), pitch(β), and Z_U .

$$l_1 = \sqrt{(rC_\alpha C_\beta + X_U - R)^2 + (rS_\alpha C_\beta + Y_U)^2 + (-rS_\alpha Z_U)^2} \quad (19)$$

$$l_2 = \sqrt{\left(\frac{\sqrt{3}r(C_\alpha S_\beta S_\gamma - S_\alpha C_\gamma) - rC_\alpha C_\beta + 2X_U + R}{2}\right)^2 + \left(\frac{2Y_U - rS_\alpha C_\beta + \sqrt{3}r(S_\alpha S_\beta S_\gamma + C_\alpha C_\gamma) - \sqrt{3}R}{2}\right)^2 + \left(\frac{rS_\beta + \sqrt{3}r(C_\beta S_\gamma) + 2Z_U}{2}\right)^2} \quad (20)$$

$$l_3 = \sqrt{\left(\frac{2X_U + R - rC_\alpha C_\beta - \sqrt{3}r(C_\alpha S_\beta S_\gamma - S_\alpha C_\gamma)}{2}\right)^2 + \left(\frac{2Y_U - rS_\alpha C_\beta - \sqrt{3}r(S_\alpha S_\beta S_\gamma + C_\alpha C_\gamma) + \sqrt{3}R}{2}\right)^2 + \left(\frac{rS_\beta + \sqrt{3}r(C_\beta S_\gamma) + 2Z_U}{2}\right)^2} \quad (21)$$

The kinematic architecture of the parallel manipulator is constructed based on the closed-loop kinematic chains as illustrated in Figure 10A. In order to solve the forward kinematic problem for the manipulator based on the position variation of each actuator as a function of time, three vector loop equations are derived by summing up the vectors presented in Figure 10 and expressed in Eq (22). Similarly, the inverse kinematics of the manipulator is formulated by rearranging the vector loop Eqs (23-25) to determine the instantaneous actuator position required for the end effector to reach the desired position.

$$X_{Ui}|_{n1} + Y_{Ui}|_{n2} + Z_{Ui}|_{n3} - R|R_0^{Ai}(1 : 3, 1) - l_i|R_0^{Bi}(1 : 3, 1) = 0 \quad (22)$$

$$X_{U1}|_{n1} + Y_{U1}|_{n2} + Z_{U1}|_{n3} - R|R_0^{A1}(1:3,1) - l_1|R_0^{B1}(1:3,1) = 0 \quad (23)$$

$$X_{U2}|_{n1} + Y_{U2}|_{n2} + Z_{U2}|_{n3} - R|R_0^{A2}(1:3,1) - l_2|R_0^{B2}(1:3,1) = 0 \quad (24)$$

$$X_{U3}|_{n1} + Y_{U3}|_{n2} + Z_{U3}|_{n3} - R|R_0^{A3}(1:3,1) - l_3|R_0^{B3}(1:3,1) = 0 \quad (25)$$

where rotational matrices;

$$R_0^{A1} = \begin{bmatrix} 1 & 0 & 0 \\ 0 & 1 & 0 \\ 0 & 0 & 1 \end{bmatrix}, R_0^{B1} = R_0^{A1} R_{A1}^{B1} = \begin{bmatrix} -\cos \theta_1 & 0 & \sin \theta_1 \\ 0 & 1 & 0 \\ \sin \theta_1 & 0 & \cos \theta_1 \end{bmatrix} \quad (26)$$

$$R_0^{A2} = \begin{bmatrix} -\cos 60^\circ & \sin 60^\circ & 0 \\ \sin 60^\circ & -\cos 60^\circ & 0 \\ 0 & 0 & 1 \end{bmatrix}, R_0^{B2} = R_0^{A2} R_{A2}^{B2} = \begin{bmatrix} \frac{\cos \theta_2}{2} & \frac{\sqrt{3}}{2} & -\frac{\sin \theta_2}{2} \\ -\frac{\sqrt{3}}{2} \cos \theta_2 & \frac{-1}{2} & \frac{\sqrt{3}}{2} \sin \theta_2 \\ \sin \theta_2 & 0 & \cos \theta_2 \end{bmatrix} \quad (27)$$

$$R_0^{A3} = \begin{bmatrix} -\cos 60^\circ & \sin 60^\circ & 0 \\ -\sin 60^\circ & \cos 60^\circ & 0 \\ 0 & 0 & 1 \end{bmatrix}, R_0^{B3} = R_0^{A3} R_{A3}^{B3} = \begin{bmatrix} \frac{\cos \theta_3}{2} & \frac{\sqrt{3}}{2} & -\frac{\sin \theta_3}{2} \\ -\frac{\sqrt{3}}{2} \cos \theta_3 & \frac{-1}{2} & \frac{\sqrt{3}}{2} \sin \theta_3 \\ \sin \theta_3 & 0 & \cos \theta_3 \end{bmatrix} \quad (28)$$

After the rearrangement of the Eqs (23-28), the coordinate of each linear actuator in **u** frame is expressed by Eqs (29-31).

$$X_{U1} = R - l_1 \cos(\theta_1), Y_{U1} = 0, Z_{U1} = l_1 \sin(\theta_1) \quad (29)$$

$$X_{U2} = \frac{-R + l_2 \cos(\theta_2)}{2}, Y_{U2} = \frac{\sqrt{3}(R - l_2 \cos(\theta_2))}{2}, Z_{U2} = l_2 \sin(\theta_2) \quad (30)$$

$$X_{U3} = \frac{-R + l_3 \cos(\theta_3)}{2}, Y_{U3} = \frac{\sqrt{3}(-R + l_3 \cos(\theta_3))}{2}, Z_{U3} = l_3 \sin(\theta_3) \quad (31)$$

The positioning of the actuators on the end-effector forming the corners of the triangular shape depicted in Figure 10 enables us to calculate the coordinate of the centroid of the triangle, namely the end-effector (X_c, Y_c, Z_c) in Eq (32) by averaging the well-defined coordinates, X_U, Y_U, Z_U .

$$X_c = \frac{\sum_{i=1}^3 X_{Ui}}{3}, Y_c = \frac{\sum_{i=1}^3 Y_{Ui}}{3}, Z_c = \frac{\sum_{i=1}^3 Z_{Ui}}{3} \quad (32)$$

4.1.2 Dynamic Analysis of the Manipulator

The 3-DoF parallel manipulator considered in this work is equipped with three linear actuators mounted between the end-effector and the base of the robot. The end-effector is able to rotate in roll (α) and pitch (β) angles and moves in the z-axis. The dynamic model of a 3-DoF parallel manipulator in joint space is derived utilizing the Euler-Lagrange method. Accordingly, the solutions extract the required forces/torques that maintain the system to follow the reference trajectory.

The general form of the dynamic equation is,

$$M(X)\ddot{X} + C(X, \dot{X}) + G = F \quad (33)$$

where X symbolizes the generalized coordinate vector of the manipulator, M represents the mass vector, C is the centrifugal and Coriolis matrix, G is the gravity vector, and F is a vectorial representation of the required forces to actuating each linear motor in task space.

The kinematics of the platform is structured based on the unit vector ($\hat{s}_i$) along the actuator direction utilizing the loop equations discussed in the kinematics analysis of the manipulator.

To find the dynamic equation, the closed-loop equation, and link lengths are redefined by using unit vector ($\hat{s}_i$) along the actuator direction. $\hat{s}_1$, $\hat{s}_2$, and $\hat{s}_3$ are unit vectors are in the same direction as axes K1, M1, and P1 shown in Figure 10, respectively

$$X_c = A_i + l_i \hat{s}_i - B_i \quad (34)$$

As found in inverse kinematics, the length of each actuator found as

$$l_i = || X_c + B_i - A_i ||_2 \quad (35)$$

To express the length of the actuators in the coordinate frame, unit vectors are defined

as,

$$\widehat{s}_i = \frac{X_c + B_i - A_i}{l_i} \quad (36)$$

An intermediate variable, δ_i , is assigned as the position of the B_i points represented in Figure 10 and derived as a means of vector summations in two different vector loop structures as shown in Figure 11B,

$$\delta_i = X_c + B_i \quad (37)$$

$$\delta_i = A_i + l_i \widehat{s}_i \quad (38)$$

By differentiation Eq (37) and Eq (38), velocities of intermediate variable can be found as,

$$\dot{\delta}_i = \dot{X}_c + w \times B_i \quad (39)$$

$$\dot{\delta}_i = \dot{l}_i \widehat{s}_i + l_i w_i \times \widehat{s}_i \quad (40)$$

where w_i and $\dot{l}_i$ represent the angular velocity and length rate of each actuator, respectively. Symbol w is the angular velocity of the end effector. Again, by differentiation Eq (39) and Eq (40), acceleration of intermediate variable can be found as,

$$\ddot{\delta}_i = \ddot{X}_c + w \times B_i + w \times (w \times B_i) \quad (41)$$

$$\ddot{\delta}_i = \ddot{l}_i \widehat{s}_i + l_i w_i \times \widehat{s}_i + \dot{l}_i w_i \times \widehat{s}_i + l_i \dot{w}_i \times \widehat{s}_i + l_i w_i \times (w_i \times \widehat{s}_i) \quad (42)$$

Then by applying dot product between Eq (39) and $\widehat{s}_i$, angular and linear velocity of the intermediate variable can express with the position of the actuators,

$$\dot{l}_i = \dot{\delta}_i \widehat{s}_i = [\dot{X}_c + w \times B_i]^T \cdot s_i \quad (43)$$

Applying cross product procedure in Eq (39) and Eq (40) to $\hat{s}_i$, angular velocity of each actuator yield,

$$w_i = \frac{\hat{s}_i \times \dot{\delta}_i}{l_i} = \frac{\hat{s}_i \times (\dot{X}_c + w \times B_i)}{l_i} \quad (44)$$

In order to make calculations easier, dynamic analysis of the actuators and moving platforms are done separately and later combined with the Jacobian matrix.

4.1.2.1 Dynamic Analysis of Each Actuation Unit

The dynamic equation in Eq (33) stated in task space is reformed and expressed as Eq (45) to find the dynamic equation of each actuator in joint space.

$$M_i \ddot{\delta}_i + C_i \dot{\delta}_i + G_i = F_i \quad (45)$$

where M_i , C_i , G_i , and F_i represent mass matrix, centrifugal and Coriolis matrix, gravity vector of the actuators, and generated force by each actuator, respectively.

The assembly of the linear actuator is composed of two main parts as represented Figure 11C. The stationary part is responsible for power generation, the moving part is in charge of the displacement of the shaft. Accordingly, the dynamic analysis of the overall action unit is derived considering the mechanical characteristics, e.g., mass, geometry, and CoM, of each part. The CoM with respect to its own frame and mass of stationary part and moving part are represented by symbols c_{i1} , c_{i2} and m_{i1} , m_{i2} , respectively.

$$\vec{c}_{i1} = A_i + c_{i1} \hat{s}_i \quad (46)$$

$$\vec{c}_{i2} = A_i + (l_i - c_{i2}) \hat{s}_i \quad (47)$$

The velocities of the stationary and moving parts are obtained by the time derivative of Eq (46) and Eq (47) and expressed in Eq (48) and Eq (49).

$$\vec{v}_{i1} = c_{i1} (w_i \times \hat{s}_i) \quad (48)$$

$$\vec{v}_{i2} = \dot{l}_i \hat{s}_i + (l_i - c_{i2}) (w_i \times \hat{s}_i) \quad (49)$$

The skew-symmetric matrix form of a vector is represented by the symbol $S(\cdot)$. Accordingly, Eqs (44, 48 and 49) are reconstructed in the form of $S(\cdot)$ and expressed in Eqs (50, 51, 52 and 53).

$$w_i = \frac{S(\hat{s}_i) \dot{\delta}_i}{l_i} \quad (50)$$

$$\dot{w}_i = \frac{S(\hat{s}_i) \dot{\delta}_i - 2l_i \dot{w}_i}{l_i} \quad (51)$$

$$\vec{v}_{i1} = \frac{-c_{i1}}{l_i} (S(\hat{s}_i)^2 \dot{\delta}_i) \quad (52)$$

$$\vec{v}_{i2} = \left(\frac{l_i - c_{i2}}{l_i} S(\hat{s}_i)^2 + \hat{s}_i \hat{s}_i^T \right) \dot{\delta}_i \quad (53)$$

Based on the Euler-Lagrange formalism, the kinetic energy of the linear actuator is formed as in Eq (54).

$$K_i = \frac{1}{2} \dot{\delta}_i^T M_i \dot{\delta}_i = \frac{1}{2} \vec{v}_{i1}^T m_{i1} \vec{v}_{i1} + \frac{1}{2} \vec{v}_{i2}^T m_{i2} \vec{v}_{i2} + \frac{1}{2} \dot{w}_i^T (I_{c_{i1}} + I_{c_{i2}}) \dot{w}_i \quad (54)$$

where $I_{c_{i1}}$ and $I_{c_{i2}}$ represent the moment of inertia (MoI) of the stationary and moving part, respectively.

Rearranging the Eq (54) yields the kinetic energy of the actuation unit expressed in Eq (55).

$$K_i = \frac{\dot{\delta}_i^T}{2} [(m_{i1} c_{i1}^2 + m_{i2} (l_i - c_{i2})^2) S(\hat{s}_i)^2 + m_{i2} \hat{s}_i \hat{s}_i^T - \frac{S(\hat{s}_i) (I_{c_{i1}} + I_{c_{i2}}) S(\hat{s}_i)}{l_i^2}] \dot{\delta}_i \quad (55)$$

The mass matrix, M_i is extracted from the kinetic energy Eq (55) and expressed in Eq (56).

$$M_i = (m_{i1} c_{i1}^2 + m_{i2} (l_i - c_{i2})^2) S(\hat{s}_i)^2 + m_{i2} \hat{s}_i \hat{s}_i^T - \frac{S(\hat{s}_i) (I_{c_{i1}} + I_{c_{i2}}) S(\hat{s}_i)}{l_i^2} \quad (56)$$

Deriving the potential energy equation in Eq (57) of the actuation unit is also required to generate the analysis of the dynamics.

$$P_i = -g^T (m_{i1}c_{i1}\hat{s}_i + m_{i2}(l_i - c_{i2})\hat{s}_i) \quad (57)$$

, where g is gravity constant, $g = [0; 0; 9.8]$.

The gravity matrix is generated as in Eq (58) by deriving the potential energy equation with respect to the variable, δ_i .

$$G_i = \frac{\partial P_i}{\partial \delta_i} = \left[\frac{(m_{i1}c_{i1}^2 + m_{i2}(l_i - c_{i2})^2)S(\hat{s}_i)^2}{l_i} - m_{i2}\hat{s}_i \cdot \hat{s}_i^T \right] g \quad (58)$$

Centrifugal and Coriolis force matrix is derived utilizing the Euler-Lagrange formalism in Eq (59) and detailed in Eq (60).

$$C_i \delta_i = \dot{M}_i \dot{\delta}_i - \frac{\partial(\dot{\delta}_i^T M_i \dot{\delta}_i)}{2\partial \delta_i} \quad (59)$$

$$C_i = \frac{-m_{i2}c_{i2}\hat{s}_i \dot{\delta}_i^T S(\hat{s}_i)^2}{l_i^2} - \frac{w_i \hat{s}_i^T (I_{c_{i1}} + I_{c_{i2}}) S(\hat{s}_i)}{l_i^2} \dots$$

$$+ \frac{2\dot{l}_i}{l_i^3} ((m_{i1}c_{i1}^2 + m_{i2}(l_i - c_{i2})^2 - m_{i2}l_i(l_i - c_{i2}))S(\hat{s}_i)^2 + S(\hat{s}_i)(I_{c_{i1}} + I_{c_{i2}})S(\hat{s}_i)) \quad (60)$$

4.1.2.2 Dynamic Analysis of End-Effector

The general dynamic equation of the end-effector of the robotic platform, the upper and lower platforms, is expressed in Eq (61).

$$M_e(X)\ddot{X} + C_e(X, \dot{X}) + G_e = F_e \quad (61)$$

, where M_e , C_e , G_e , and F_e represent mass matrix, centrifugal and Coriolis matrix, gravity vector of the end-effector, and force exerted by the end-effector, respectively.

The position and orientation of the end-effector with respect to the base frame are expressed by $X = [x; y; z; \alpha; \beta; \gamma]$. The time derivative of X that constitutes the velocity of the end-effector yields $\dot{X} = [\dot{x}; \dot{y}; \dot{z}; \dot{\alpha}; \dot{\beta}; \dot{\gamma}]$.

The dynamic analysis of the end-effector requires transforming the variables from the base frame to their own frame through the R^T matrix.

$$R^T = R_x(\alpha)^T R_y(\beta)^T R_z(\gamma)^T = \begin{bmatrix} C_\gamma C_\beta & C_\beta S_\gamma & -S_\beta \\ C_\gamma S_\alpha S_\beta - C_\alpha S_\gamma & S_\gamma S_\alpha S_\beta + C_\alpha C_\gamma & S_\alpha C_\beta \\ C_\gamma C_\alpha S_\beta + S_\alpha S_\gamma & S_\gamma C_\alpha S_\beta - S_\alpha C_\gamma & C_\alpha C_\beta \end{bmatrix} \quad (62)$$

The angular velocity of the end-effector with respect to its own frame is expressed utilizing the Eq (62).

$$w = R^T \begin{bmatrix} \alpha \\ \beta \\ \gamma \end{bmatrix} = \begin{bmatrix} C_\gamma C_\beta & C_\beta S_\gamma & -S_\beta \\ C_\gamma S_\alpha S_\beta - C_\alpha S_\gamma & S_\gamma S_\alpha S_\beta + C_\alpha C_\gamma & S_\alpha C_\beta \\ C_\gamma C_\alpha S_\beta + S_\alpha S_\gamma & S_\gamma C_\alpha S_\beta - S_\alpha C_\gamma & C_\alpha C_\beta \end{bmatrix} \begin{bmatrix} \alpha \\ \beta \\ \gamma \end{bmatrix} \quad (63)$$

The angular acceleration of the end-effector with respect to its own frame is derived by means of R^T and its time derivative, $\dot{R}^T$, and expressed in Eq (64).

$$\dot{w} = \dot{R}^T \begin{bmatrix} \dot{\alpha} \\ \dot{\beta} \\ \dot{\gamma} \end{bmatrix} + R^T \begin{bmatrix} \ddot{\alpha} \\ \ddot{\beta} \\ \ddot{\gamma} \end{bmatrix} \quad (64)$$

The kinetic energy equation of the end-effector to be employed in the Euler-Lagrange formalism is formed as in Eq (65).

$$K_e = \frac{1}{2} \dot{x}_e^T M_e \dot{x}_e = \frac{1}{2} v_e^T m_e v_e + \frac{1}{2} w^T I_e w \quad (65)$$

, where m_e and I_e are the mass and inertia matrix of the end-effector. Here, v_e is the time derivative of Eq (32).

The mass matrix, M_e is extracted from the kinetic energy Eq (65) and expressed in

$$M_e = \begin{bmatrix} m_e I_{3 \times 3} & 0 \\ 0 & R I_e R^T \end{bmatrix} \quad (66)$$

The potential energy equation in Eq (67) of the end-effector is needed to analyze the dynamics of the end-effector.

$$P_e = -g \begin{bmatrix} -m_e X_c \\ 0 \\ 0 \end{bmatrix} \quad (67)$$

The gravity matrix is generated as in Eq (68) by deriving the potential energy function with respect to the vector X .

$$G_e = \frac{\partial P_e}{\partial X_c} = \begin{bmatrix} -m_e g \\ 0 \end{bmatrix} \quad (68)$$

Centrifugal and Coriolis force matrix is derived utilizing the Euler-Lagrange formula in Eq (69).

$$C_e = \begin{bmatrix} 0 & 0 \\ 0 & \dot{R}I_e R^T + RS(w)I_e R^T \end{bmatrix} \quad (69)$$

4.1.2.3 Dynamic Analysis of the Overall System

The intermediate variable (δ_i) configured by each linear actuator affects the position and orientation of the end-effector in task space in an independent manner. Considering the characteristics of the parallel manipulator, the configuration of each actuation unit affects the position and orientation of the end-effector's frame. Since the dynamic representation of the robotic platform, (M_{all} , C_{all} , and G_{all}), needs to be expressed in task space, the dynamic equation of each linear actuator written with respect to its own frame as expressed in Section 4.1.1 has to be stated in task space through the Jacobian matrix in Eq (66).

$$J_i = \begin{bmatrix} I_{3 \times 3} & -S(B_i)T_{reverse} \end{bmatrix} \quad (70)$$

The dynamic contributions of both the linear actuators and end-effector are added together as in Eq 71, 72, and 73.

$$M_{all} = M_e + \sum_{i=1}^3 J_i^T M_i J_i \quad (71)$$

$$C_{all} = C_e + \sum_{i=1}^3 (J_i^T M_i J_i + J_i^T C_i J_i) \quad (72)$$

$$G_{all} = G_e + \sum_{i=1}^3 J_i^T G_i \quad (73)$$

Considering the dynamic effect of each member of the system, the generated force to actuate the end-effector in the workspace is calculated by the Eq (74).

$$M_{all}(X)\ddot{X} + C_{all}(X, \dot{X}) + G_{all} = F_{all} \quad (74)$$

4.1.3 Simulation of the kinematics and dynamics of the 3 DoF parallel manipulator

In this Section, the derivations of kinematics and dynamics of the 3 DoF parallel manipulator are mapped in the simulation environment. The overall flowchart of the simulation to represent forward and inverse kinematics and dynamics obtained in Section '*Kinematic Analysis of the Manipulator*' and '*Dynamic Analysis of the Manipulator*' is presented in Figure 12.

The reference trajectory given at the beginning of the simulations is compared with the reference trajectory obtained as a result of inverse dynamics. In summary, the state variables, roll(α) and pitch(β) angles, and z translation of the end-effector (upper and lower platforms) are derived to be employed in the position, velocity, and acceleration level kinematics since only these three DoFs are controllable. Then, the constraint Eqs (15-17) are utilized to obtain the other three dependent equalities of yaw(γ), x , and y , and their derivatives are operated as the reference signals in velocity and acceleration levels. The reference variables defined in task space are

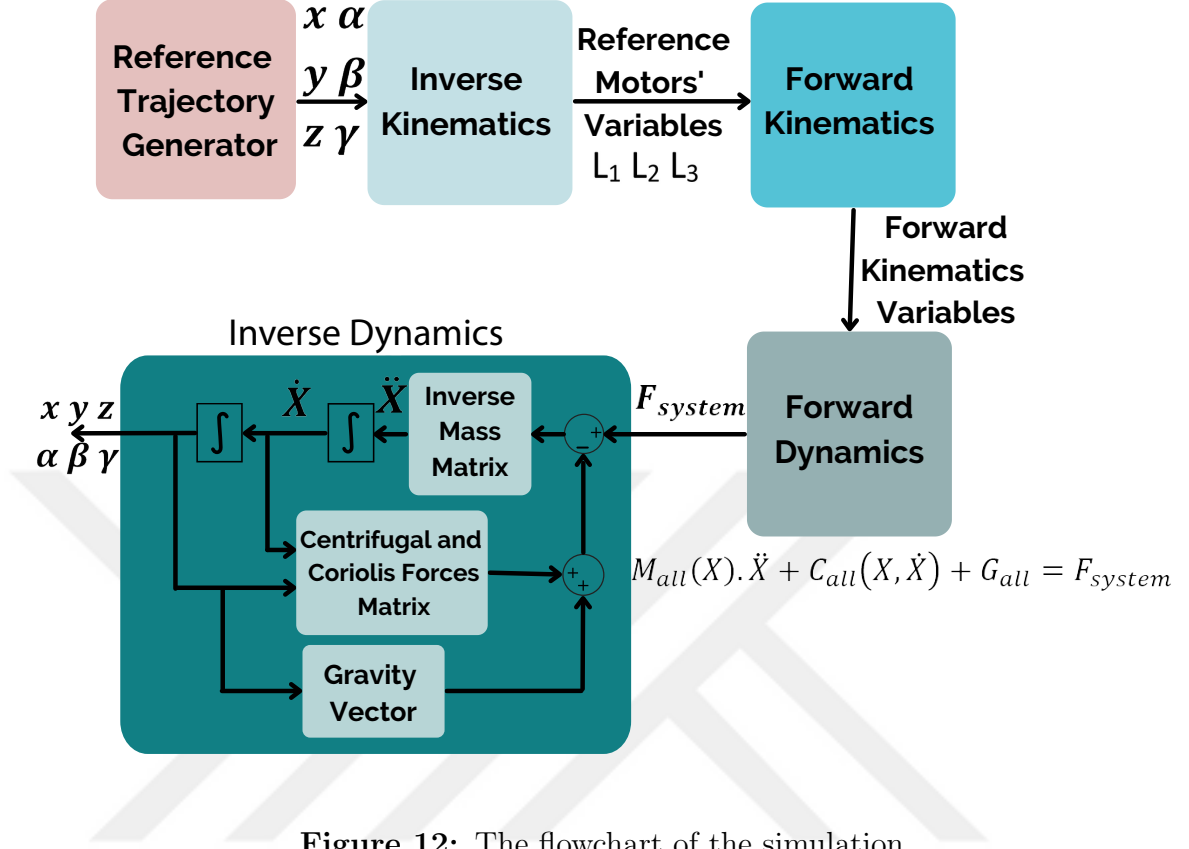


Figure 12: The flowchart of the simulation

represented in joint space with the use of inverse kinematic equations, e.g., required reference motor variables are found from reference end-effector orientation. Then, the forward kinematic representations are utilized to obtain force and torque equalities by Eqs (67, 68, and 69). Lastly, the inverse dynamic analysis computes the states of the end-effector (position, velocity, and acceleration) based on torques and forces generated by the actuators.

The simulation results indicate that the actual trajectories of the end-effector in Cartesian coordinate space are well-fitted with the reference trajectories. The Figure 13A shows that the translational position of the end-effector follows the reference trajectory well with Root-Mean-Square Errors (RMSEs) 4.304×10^{-7} mm in the x direction, 4.3×10^{-8} mm in the y direction, 4.284×10^{-7} mm in the z-direction. Similarly, Figure 13B provides evidence that the reference trajectory is well-matched with the rotational position of the end effector. Accordingly, the tracking errors in

roll, pitch, and yaw directions are 2.824×10^{-7} rad, 1.316×10^{-6} rad, and 7.224×10^{-7} rad, respectively. Towards this end, the simulation results validate the accuracy of the derivations of kinematics and dynamics of the system.

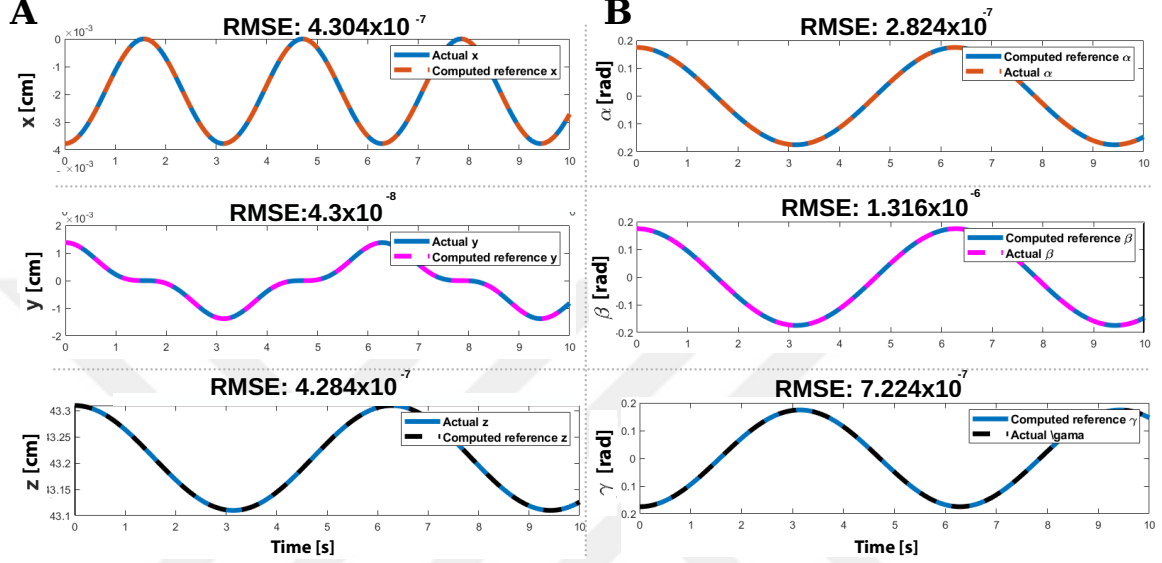


Figure 13: Simulation results (A) Translational motion of the end-effector (B) Rotational motion of the end-effector

4.2 ADAMS Analysis

The linear actuator is selected according to the obtained force/torque values in the simulation. As stated earlier, the parallel manipulator has 3-DoF (rotations in roll and pitch directions and linear translation in the z-axis) and includes three linear actuators (see Figure 14A). Each actuator acts in accordance with different references after controllable DoFs (rotation in roll and pitch and z translation) is given as a reference constraint equations are used to solve the yaw rotation and x and y translation; the detailed structure is expressed in the previous Section. Therefore, the positions of the motor movements to be given in the simulation are found by using inverse kinematics equations in MATLAB, and then the obtained function is written in the ADAMS program. Two motion is shown as an example; the first one is for the solely z-translation in this motion, all actuators are moving in the same direction, but the

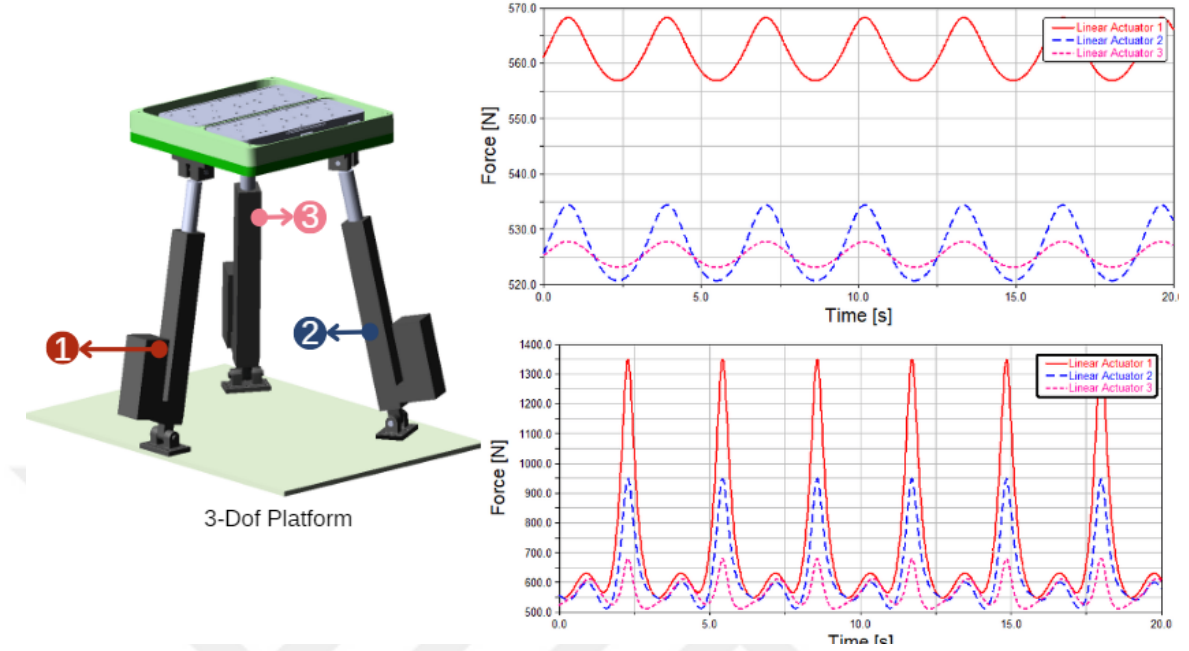


Figure 14: (A) ADAMS model, (B) Required forces at z translation and (C) at circle motion

maximum force is found as 568 N in the first linear actuator (end effector is not symmetrical thus load distribution creates this difference), as depicted in Figure 14B. Similarly, second motion for the circular motion of the end effector in this motion all actuators work against each other so the force requirement is highest. The first linear actuator demonstrates a maximum required force of 1350 N, as shown in Figure 14C, and servo motor capacity is chosen accordingly.

4.3 ANSYS Analysis

The feasibility of the designed system, the safety of the person on the system, and the reaction of the system under static stress are evaluated with the ANSYS program. Thus, the stress, displacement, strain, and durability of the system are tested.

The material selection criterion is stiffness, strength, toughness, lightweight, and cost (see Figure 15A). The weight of the patient is distributed to foot plates, load cells, and lower platform, so these parts must have a high young modulus. Among the metallic materials considered, aluminum emerges as the optimal choice due to its

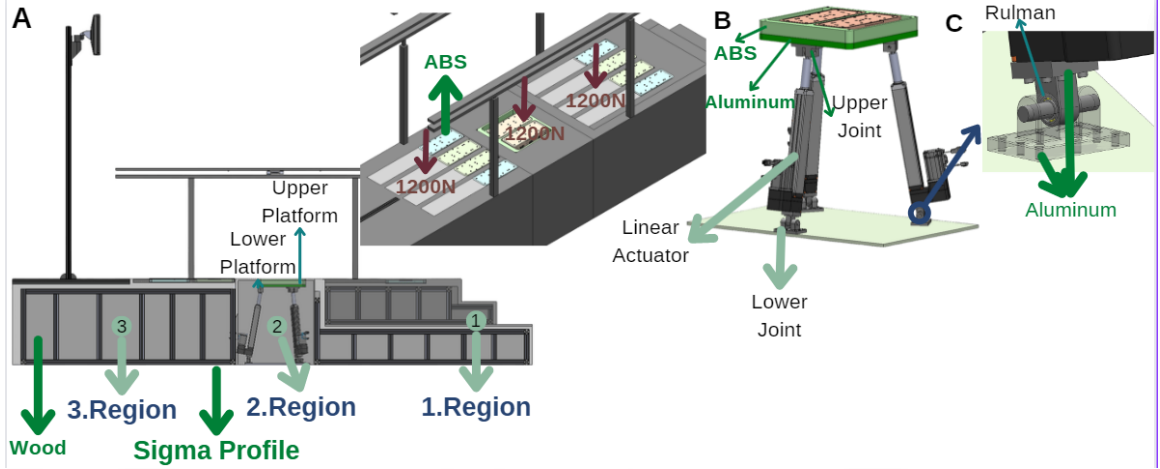


Figure 15: Applied force and material selection on the (A) system, (B) 3 DoF parallel manipulator, and (C) Lower joint

lightweight nature and high modulus, ensuring both durability and minimal strain on the linear actuator. Furthermore, the material selection for this component is made as ABS to prevent direct force application by the patient on the upper platform frame. In addition, wood is chosen as the material for the outer frames, while sigma profiles are incorporated within the frames.

The maximum results obtained indicate that the compression force results in a deformation of less than 1 mm, which is considered negligible and confirms the rigid and durable nature of the design. Thus, the finite element analysis conducted further reinforces the system's adherence to safety criteria.

4.4 *Simscape Analysis*

The dynamic model of the 3-DoF parallel manipulator is composed of nonlinear functions of the state variables, i.e., mass matrix, centrifugal and coriolis matrix, gravity vector, forward and inverse kinematics which are explained in the previous Section. This characteristic of the dynamic model makes the closed-loop control system non-linear and difficult to solve. Computed torque control (CTC) is specifically useful for

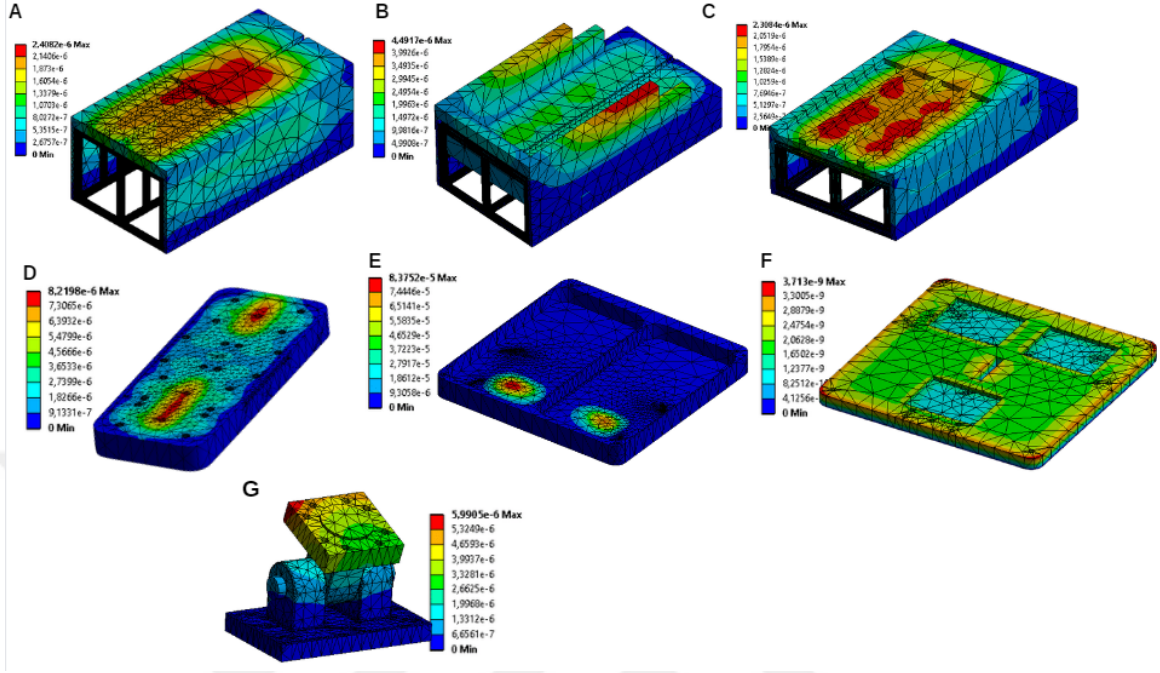


Figure 16: Deformation of (A) 3rd region, (B) 2nd region (C) 1st region, (D) Foot-plate, (E) upper platform, (F) Lower platform, (G) lower joint

parallel manipulator types since its additional second loop leads to feedback linearization. Thus, the Simscape dynamic simulation is conducted to verify the behavior of both the selected material and linear actuators. The proposed parallel manipulator is modeled using the simmechanics multibody library in MATLAB/Simulink. The CTC of linear actuators in a dynamic environment is evaluated by this simulation (see. Figure 17 and Figure 18).

The system is a parallel manipulator with three links, actuated with three linear actuators (see. Figure 18B that are placed in green (first), blue (second), and yellow (third) colored areas) between the base and the end effector. To construct this model in Simulink, the initial steps involve incorporating Simscape world, mechanical solver, and starting condition blocks. Subsequently, reference frame transformation is conducted for each interconnected block to achieve the desired configuration, and the appropriate kinematic behaviors for the interconnected blocks are specified by defining the respective joint motions.

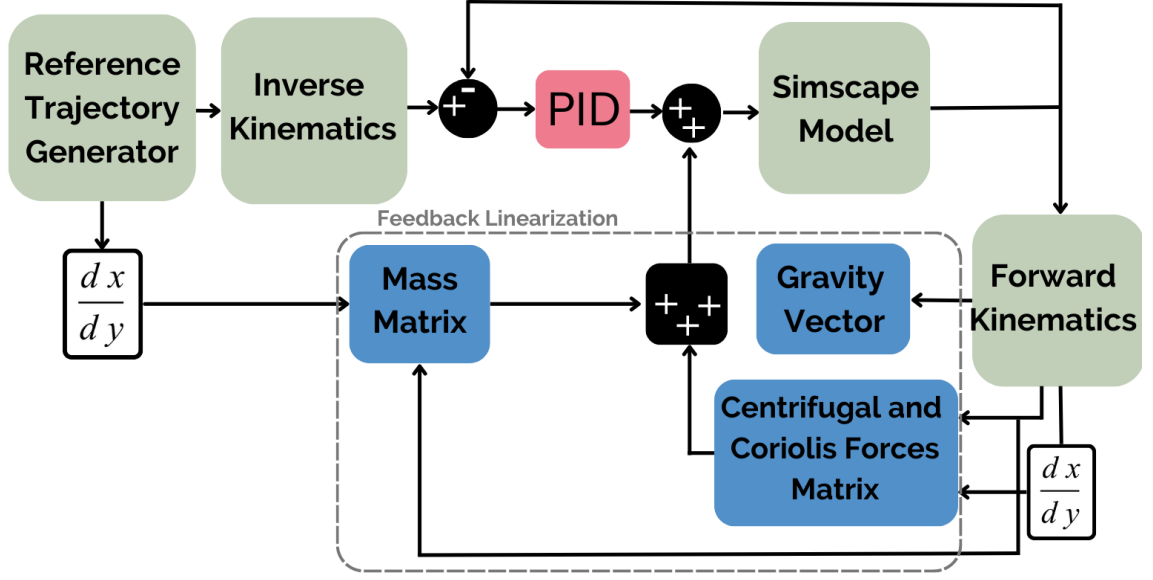


Figure 17: Computed Torque Control simulation model

Table 5: RMSE of linear actuators position tracking in Case 1: Translation in z, Case 2: Rotation in x and y, Case3: Rotation in y, and Case 4: Rotation in x

	Case 1	Case 2	Case 3	Case 4
Actuator 1	3.006×10^{-6}	3.004×10^{-6}	2.982×10^{-6}	2.998×10^{-6}
Actuator 2	2.814×10^{-6}	2.777×10^{-6}	2.982×10^{-6}	2.706×10^{-6}
Actuator 3	2.077×10^{-6}	2.123×10^{-6}	2.982×10^{-6}	2.202×10^{-6}

The physical attributes, including inertia, mass, and geometry, of each solid component are extracted from CAD model files. The components are then assembled by applying appropriate frame transformations. Prismatic joints are defined in the piston part of linear actuators and force is given as input and the position of the joint is calculated via forward kinematics. Each linear actuator moves by finding the position references according to the desired orientation of the end effector with forward kinematic equations. The end effector can rotate in roll (α) and pitch (β) angles and translation in the z-axis or combination between these axes. The RMSEs of the position tracking results of 4 different end effector's references are summarized in Table 5. Even though the RMSE values are similar to each other, the maximum RMSE found on the first actuator in z-translational motion as 3.006×10^{-6} mm.

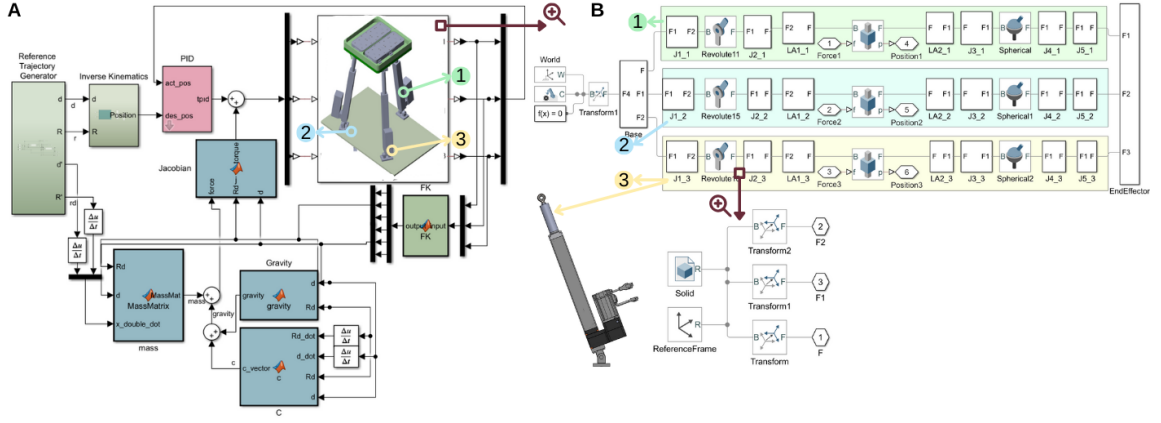


Figure 18: Computed Torque Control (A) simulation and (B) Simscape model of 3-DoF Parallel Manipulator

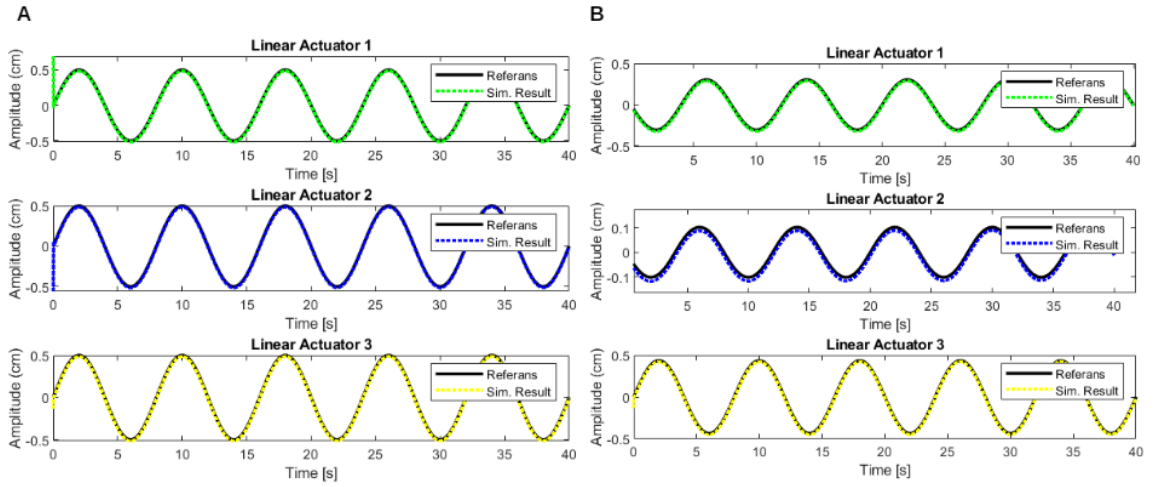


Figure 19: Linear actuator position tracking simulation result in (A) z translational motion and (B) x-y circular rotational motion of the end-effector

4.5 Discussion

Robotic rehabilitation service is presented to patients with the use of either static or dynamic platforms. The patients' balance performance on the static platform is also evaluated by ball catching and throwing exercises [145]. However, considering the validated efficacy of the PBT under the sole of the patients on motor learning [181, 182], dynamic balance systems outperform the static ones. In other words, PBT is much more effective if the bio-performance of the patient is instantaneously measured and provided as feedback. Most dynamic balance systems are not able to assess which

side of the body is ineffective during the maintenance of the balance due to the lack of a sensor fusion system embedded in the end-effector. To address the problem, Hunova [155], 2 DoF system designed for both balance and ankle rehabilitation and able to assess sole pressure of the patients, is shown the effectiveness of the system in various clinical studies [137, 183, 184, 185, 186, 187]. However, the system is limited with roll and pitch rotations and thus unable to mimic the exercises provided by physiotherapists, which purposely leads to dynamic instability in the z-axis [169] such as stepping, jumping, and stair climbing exercises.

To address the above-mentioned issues, in this thesis includes a 3-DoF parallel manipulator in the middle structure. The DoF of the manipulator covers both the ankle and balance movements in ADL, and it is aimed to improve the balance controls of the patients against the dynamic effects coming from the sole of the feet by giving perturbation in the z-axis.

The kinematic and dynamic modelings of this 3 DoF robotic platform are solved and evaluated in MATLAB/SIMULINK simulation environment based on the simulation flowchart shown in Figure 11. Considering the error values between the actual and reference trajectories, the error for pitch angle variation and the maximum RMSE among the three trajectories is found to be 1.316×10^{-6} rad. Along these lines, the results (see. Figure 13) of the simulations confirm that the kinematic and dynamic derivations of the 3 DoF robotic platform are found to be accurate.

The proposed robotic platform is evaluated in simulation before building the prototype of the system. The Adams software is employed to select the properties of the servo motors. After the model of the dynamics of the 3-DoF robotic manipulator is built, different references are given to the motors (the positions to be given are found with inverse kinematics), and the maximum required force value of the motors is found as 1350 N (see Figure 14). Furthermore, the load-bearing capacity and feasibility of the static parts of the system are analyzed with ANSYS software.

The results indicate that selected materials do not show significant deformation and are able to carry human weight (see Figure 15). The last simulation is conducted in MATLAB/Simulink; the 3-DoF parallel manipulator is modeled by using Simscape multibody library (see Figure 17). Each solid part's shape is read from the corresponding CAD model, and parts are connected with each other with their frame transformation and joint type. Then, by using the kinematic and dynamic equations, how the interaction of the selected materials and the servo motors affect the system control is investigated with the CTC method. Four different case references are given to the end effector; these are z-translation, circular motion, only pitch and only roll, respectively. The maximum RMSE is found to be 3.006×10^{-6} mm in the first servo motor in Z-transformation motion. This simulation is conducted to prove that the selected material and linear actuator can be controlled together, and non-linearity due to the parallel manipulator structure can be compensated with CTC.

CHAPTER V

DEVELOPMENT OF BALANSENS SYSTEM

This chapter presents the prototype of the 3 DoF parallel manipulator (see Figure 20 and Figure 21), and its design characteristics are evaluated during the passive ankle preparation phase. While the system comprises various components, the primary focus is on the dynamic aspect, specifically the 3-DoF parallel manipulator, because since this is a parallel manipulator, an error in the connection prevented the prototype of the entire system. In addition, the same foot plates that have been used in the parallel manipulator's end effector are also used in the stepping targets; by testing their data acquisition, a general structure can be obtained. The communication and performance characteristics of the servo motors' and load cell sensors' are investigated to improve overall system performance and enhance real-time control capabilities during the ankle preparation phase.

The robotic platform design includes a large number of sensors (load cell, vibration motor, IMU, and EMG); that's why, in the long run, in order to solve communication problems, the manipulator's sensors and motors are connected to two panels, namely the electrical and load cell panels (see Figure 22). Regarding hardware wiring, detailed attention is given to connecting various systems (servo drivers, analog and digital modules), sensors, and actuators within the system. The hardware connections have to support reliable and efficient data transfer; thus, the Ethernet for Control Automation Technology (EtherCAT) communication protocol has been implemented to contribute high-speed, real-time communication and allow for synchronized data exchange between systems by using EtherCAT Network Adapter (Crevis GN-9386). CodeSyS software is used to facilitate the coordination and exchange of data between

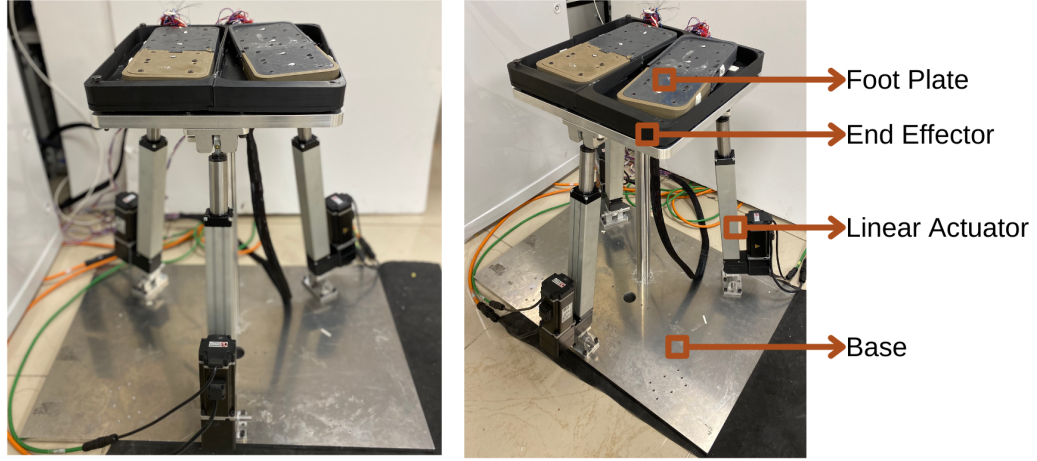


Figure 20: The prototype of the designed parallel manipulator

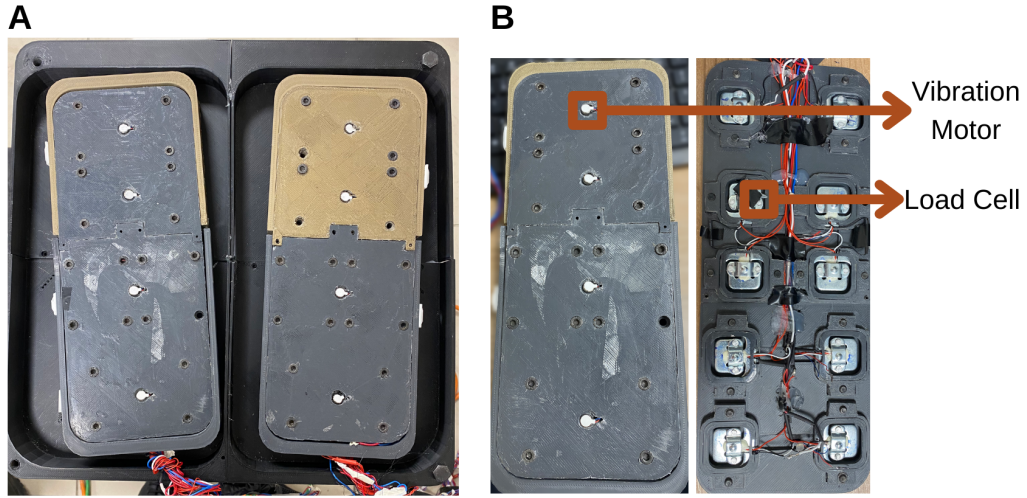


Figure 21: The prototype of (A) End effector and (B) Foot Plate

hardware modules and software applications.

The electrical panel (see Figure 23) comprises of computer (NISE 50-J1900 (Intel® Celeron® processor J1900 quad Core, 2.42GHz, 8 GB RAM)), circuit breaker, power supplies, servo driver (Estun Pronet-04AEG-EC), Solid State Relay, Terminal, Analog Input Module (CREVIS GT-3424), Digital Output Module (Crevis GT-226F), and EtherCAT Network Adapter (Crevis GN-9386). Their connection with each other is shown in Figure 23. The internal wiring of the electrical panel is made among itself, and its connection to the outside is left only with the workstation and sensor cables.

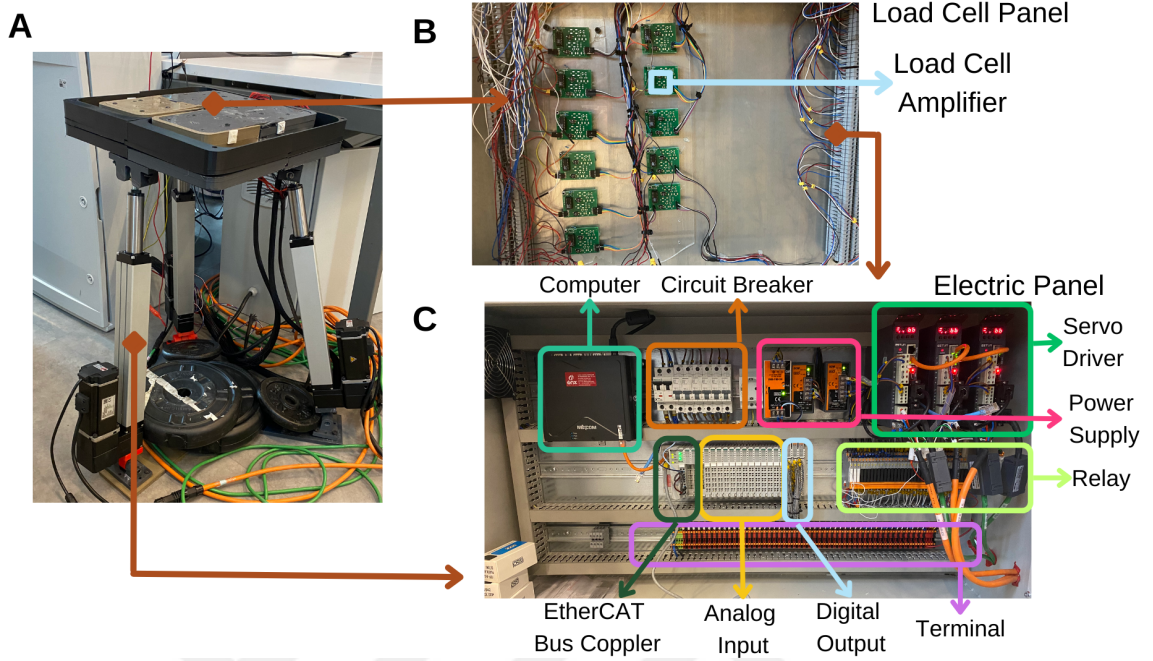
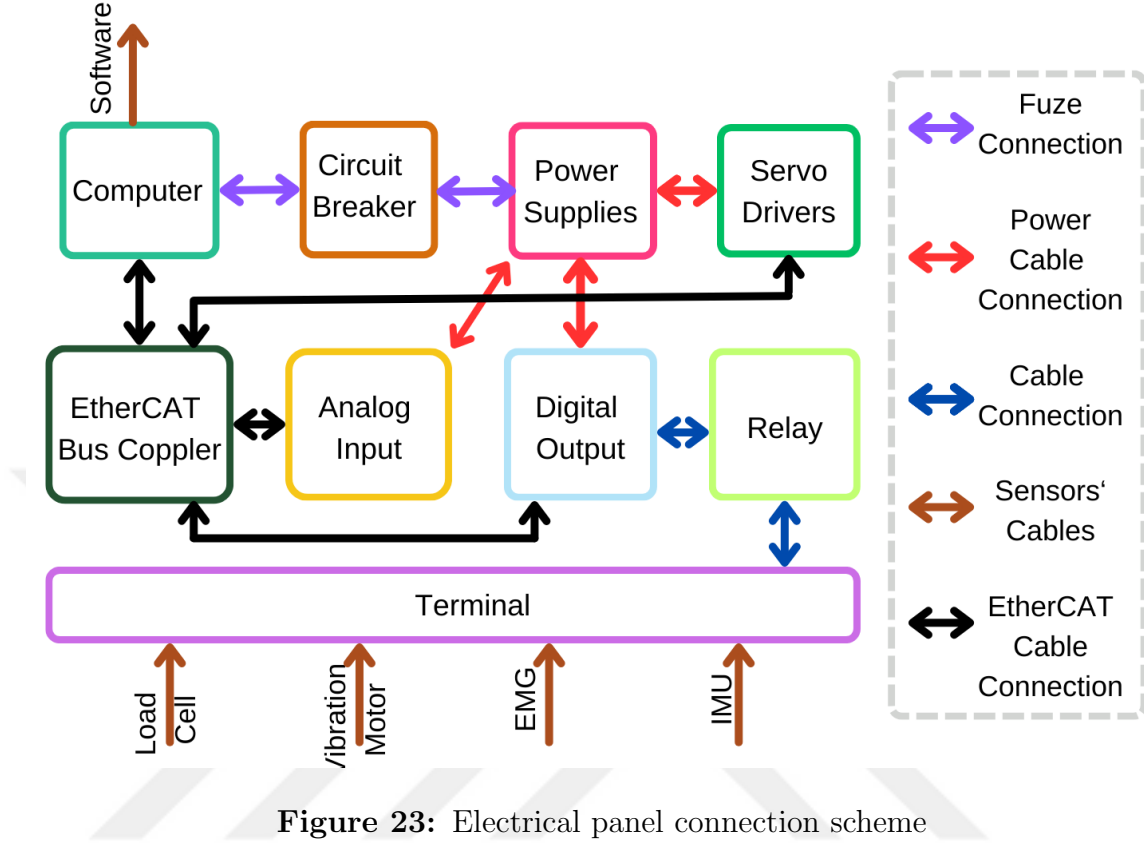


Figure 22: (A) The first prototype of the designed parallel manipulator, (B) Electrical and (C) Load cell panel

All outside sensor cables are connected to the terminal. They are not connected directly to other components because output cables can be removed easily if something goes wrong with a sensor. Load cell cables are connected to the terminals, and then corresponding cables go to analog input modules to transmit the data to the computer with EtherCAT Bus Coppler. Similarly, vibration motors terminal connections' are connected to relays, and then relays are connected to digital output modules. If the vibration motors need to be triggered according to the data received from the load cell or EMG, the digital module will trigger the relays, ensuring that they are short-circuited, thus enabling the vibration motors to become active by passing current through them.

The load cell panel only includes the designed load cell amplifier circuit, and outputs from these circuits are connected to the analog input module in the electrical panel. The amplifier circuit is designed with a notch filter to eliminate the 50 Hz power line noise. The load cell amplifier is simulated in the LtSpice software to



ensure that the notch filter can remove the 50 Hz noise signal. Then, Printed Circuit Board (PCB) is designed on the EAGLE software.

5.1 Load Cell Amplifier

Load cells are force transducers that convert force to electrical signals. They are made of strain gauges in a Wheatstone bridge position. The strain gauge's wire is designed to vary resistance when force is applied. The signal's amplitude varies in direct proportion to the force exerted. However, it is hard to detect voltage changes accurately since the change in resistance is observed in millivolts by using amplifier circuit minor differences can be magnified into more detectable signals. Load cells have four cable voltage supply (V_{input}), ground, inverting (V_{in+}) and non-inverting (V_{in-}) inputs. They are connected to the PCB and supply and ground directly connected to the integrated circuit's (IC) (INA125) supply (V_{supply}) and ground.

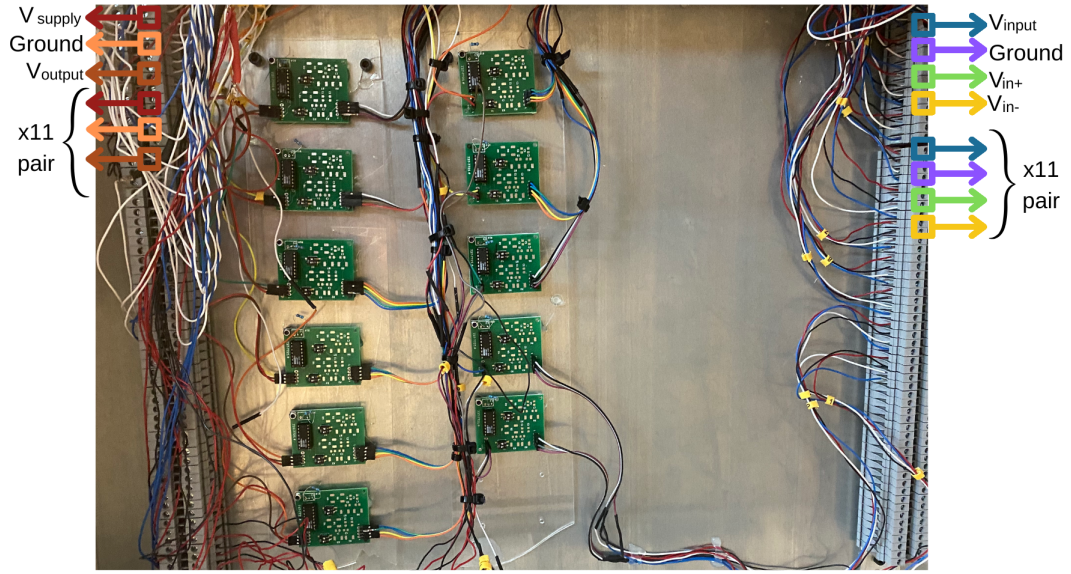


Figure 24: Load cell panel connection scheme

Similarly, inverting and non-inverting signals are connected to the notch filter then filtered signals are connected to the IC. The amplified output (V_{out}) is connected to the analog module in the electrical panel. Before this circuit (see Figure 25C) is produced, the circuit is simulated in the LtSpice program (see Figure 25A). In order to stimulate this circuit, the IC model (INA125) has been downloaded from Texas Instruments open software library and implemented into the LtSpice. After that, the required gain resistor and input signal voltage have been determined to do the simulation of the circuit. Simulation results have been shown in Figure 25B. After observing that the notch filter can minimize the noise signal in the simulation, PCB is designed accordingly (see Figure 25C).

5.2 System Characterization

The designed system includes different modules, but the main dynamic part consists of a 3-DoF parallel manipulator and sensor communication panel. Thus, their communication and characteristics are tested with several tests. The system's dynamic

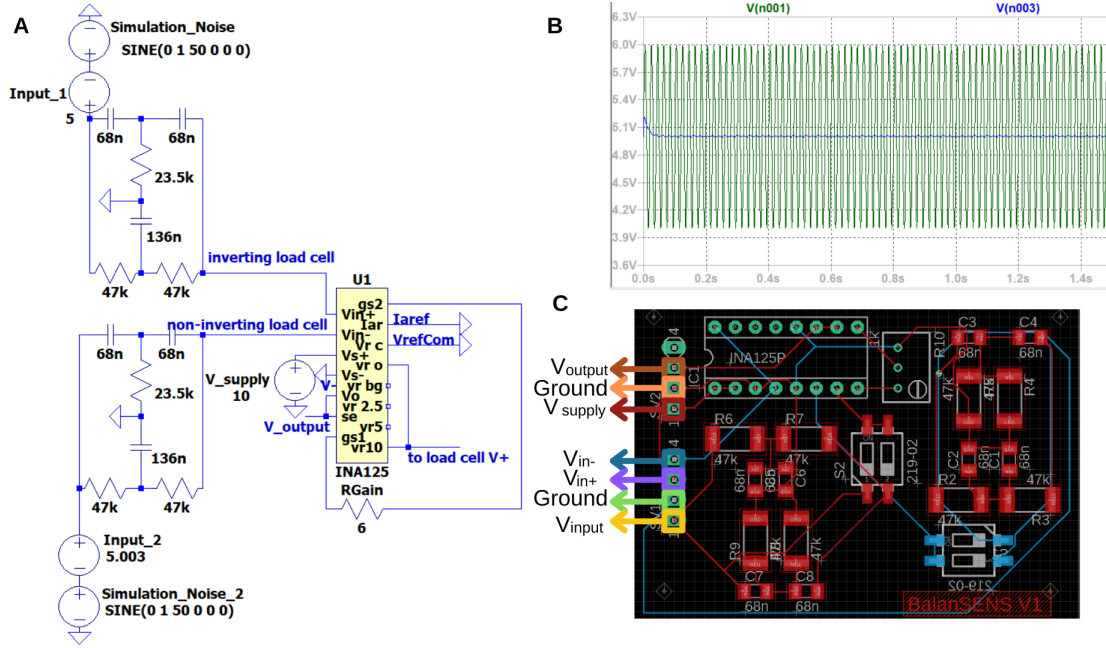


Figure 25: (A) LtSpice simulation, (B) simulation result, and (C) PCB design of load cell amplifier

behavior and the motor's performance characteristics are assessed by analyzing various frequencies in the Bode plot. To examine this, a chirp signal (initial frequency starting at 10 Hz and ending at 1 kHz for 120 seconds) is generated on the software and employed as a reference input for the servo motors (while the system is in assembly form). Subsequently, the collected data and the references are utilized in the MATLAB system identification toolbox to ascertain the corresponding transfer functions for each servo motor. The Bode plots display the gain and phase shift of the servo motors at different frequencies, thus providing information about the servo motor's performance and stability and designing appropriate control strategies.

This finding is particularly crucial in designing and implementing control algorithms, as it helps select appropriate limits in the rehabilitation. The result shows that frequencies up to 6300 Hz (see Figure 26) are the ones with high gain, and motors are controllable. The motors' responses are not controllable after this frequency, i.e., with a 3 dB drop in magnitude plot. This drop indicates a decline in the motor's

capacity to precisely and consistently react to input signals. In other words, trying to control the motor with input signals over this range may result in unpredictable or erratic behavior, making the motor more difficult to control. As shown by the magnitude plot and the point of the 3 dB drop, it is vital to stay within the frequency range where the motor's reactions are predictable and under control. The motors' responses are not controllable after this frequency, i.e., with a 3 dB drop in magnitude plot. This drop indicates a decline in the motor's capacity to precisely and consistently react to input signals. In other words, trying to control the motor with input signals over this range may result in unpredictable or erratic behavior, making the motor more difficult to control. In addition, even though the second and third servo motors have higher cutoff frequencies since they all are to be controlled with the same frequencies, the lowest one is taken as the functional point. It is vital to stay within the frequency range because, in human-subject experiments, the system must remain in control. Since all of the motors are used within the same control loop, the cutoff frequencies for the entire system are determined by the lowest value found among the motors (6300 Hz, see Figure 26). This cutoff frequency holds significant importance not only in terms of stability but also in determining the velocity of perturbation. Considering the limitation of the I-BaR framework velocities, the results derived from the characterization tests are satisfactory.

The phase plot of servo motors' shows the time delay between the input and output signals. At low frequencies, the phase shift is minimal, indicating that the motors respond fast to changes in the input signal. As the frequency increases, the phase shift becomes more significant, thus delaying the motors' response. However, it does not exceed -180 degrees thus, the open loop response of the servo motors does not operate near the instability region in the I-BaR rehabilitation requirements.

Characteristics of servo motors are identified through a series of tests by velocity, accuracy, and response time. In order to assess the velocity tests involved measuring

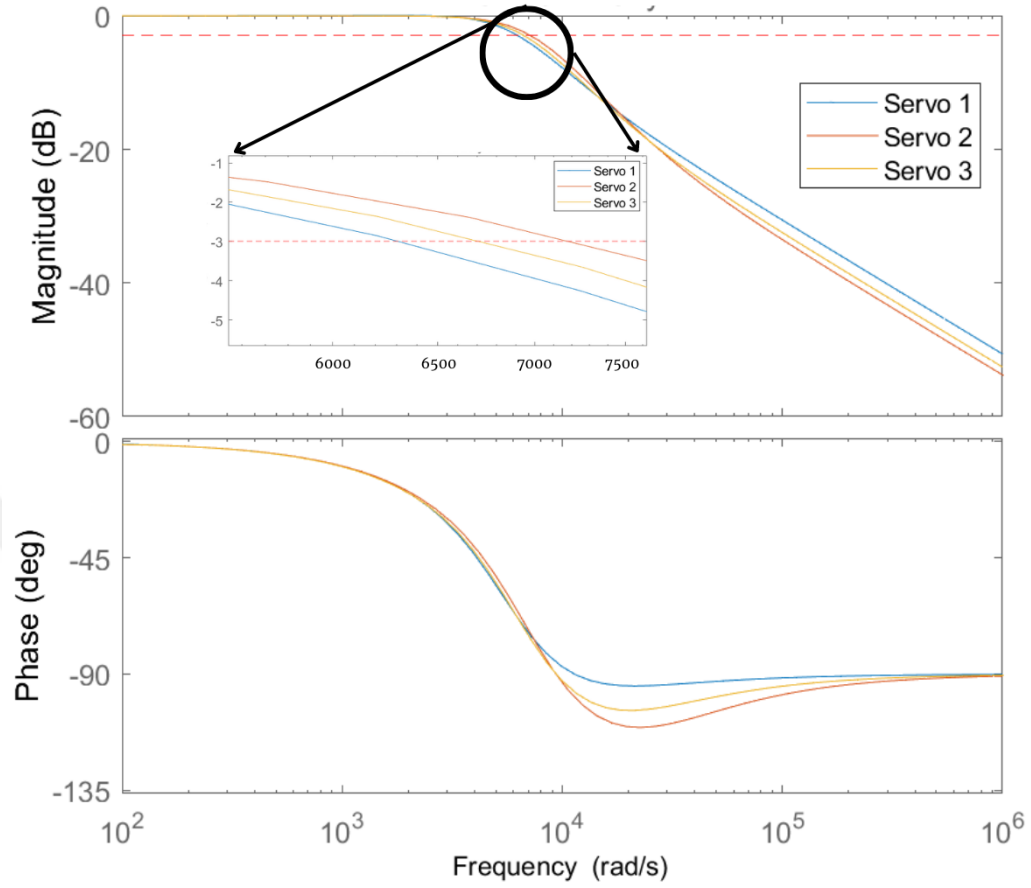


Figure 26: Frequency Response Characteristics of the Servo Motors

the motor's velocity response at different position inputs (10 to 100 mm). Accuracy tests examined the motor's ability to accurately reach and maintain specific target positions (10 to 100 mm) with the response and settling time of the servo motors. These tests provided valuable insights for the performance and behavior of the servos under different operating conditions (from 10 mm/s to 100 mm/s and 10 mm to 100 mm). By analyzing the characteristics of the servo motors, their control capacity is proven for the I-BaR application, ensuring reliable and precise control in the system.

The system is tested with different velocities (from 10 mm/s to 100 mm/s as set to be rehabilitation limit references) and target distances (from 10 mm to 100 mm is set to be the range in z translation) with a 2 ms sampling rate with five repetitions for each motion. As shown in Figure 27, in the lowest target position (10

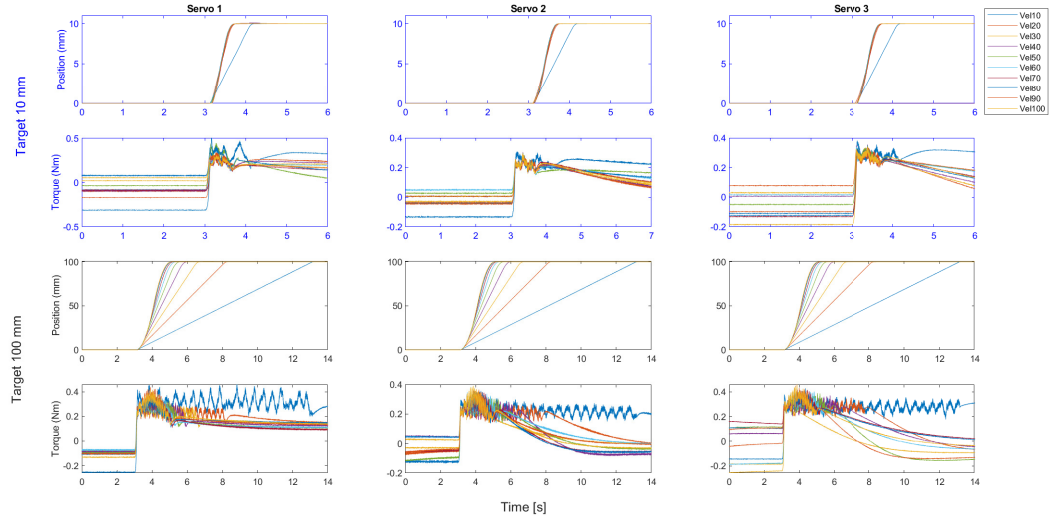


Figure 27: Servo motor position and torque tracking characteristics on low and high position target at different velocity

mm), velocities do not show a significant difference, whereas, in the highest target position (100 mm), the differences become more evident (rise time decrease, and torque response and accuracy increase) as velocity increases. Table 6 summarizes the servo motor response at its lowest and highest value. The longest rise and settling time obtained as 9.039 s and 11.261 s, 9.043 s and 10.489 s, and 9.031 s and 11.507 s for the first, second, and third servo motor, respectively, with reference to go to the 100 mm position target with 10 mm/s. Whereas the highest amount of error time is obtained as 0.773 mm, 0.772 mm, and 0.773 mm for the first, second, and third servo motor, respectively, with a reference to go to the 100 mm position target with 100 mm/s velocity.

The maximum RMSE values are obtained as 0.773 mm when servos are given 100 mm position and 100 mm/s velocity at first servo motors. This position error results in a maximum error of 0.24 degrees in the orientation of the end effector. Although this angle value does not make a significant difference for healthy people, it may be an important value in patients with high spasticity, so this value can be minimized with different control algorithm. Whereas the maximum rise and settling time are

Table 6: Rise Time, Settling Time, and RMSE for different velocity and positions

Servo motor, Reference position, and Reference velocity, respectively	Rise Time [s]	Settling Time [s]	RMSE [mm]
S1, Position 10 mm, Velocity 10 mm/s	0.942	2.172	0.110
S1, Position 10 mm, Velocity 100 mm/s	0.484	1.488	0.136
S1, Position 100 mm, Velocity 10 mm/s	9.039	11.261	0.281
S1, Position 100 mm, Velocity 100 mm/s	1.534	2.678	0.773
S2, Position 10 mm, Velocity 10 mm/s	0.942	1.750	0.100
S2, Position 10 mm, Velocity 100 mm/s	0.510	0.784	0.137
S2, Position 100 mm, Velocity 10 mm/s	9.043	10.489	0.276
S2, Position 100 mm, Velocity 100 mm/s	1.550	2.576	0.772
S3, Position 10 mm, Velocity 10 mm/s	0.932	1.992	0.107
S3, Position 10 mm, Velocity 100 mm/s	0.480	0.754	0.135
S3, Position 100 mm, Velocity 10 mm/s	9.031	11.507	0.277
S3, Position 100 mm, Velocity 100 mm/s	1.534	2.634	0.773

obtained when servos are given a 100 mm and 10 mm/s. While this characteristics test provides valuable insights into the performance of the motors, it also revealed certain limitations that should be acknowledged during ankle rehabilitation. One limitation is the presence of RMSE values for servo motor position tracking capacities ($1 \mu\text{m}$). This error is due to the non-linear kinematic equations and mechanical imperfections. Such problems are tackled by utilizing CTC to alleviate non-linearity and replacing the ABS material used for the joints with aluminum, respectively. Moreover, the experiment identifies that the maximum rise and settling times occurred when the motors are operated at a position of 100 mm and a relatively lower velocity of 10 mm/s. Additionally, the observed higher torque requirements time during this motion can cause mechanical stress on the system, which could impact its overall longevity and reliability. However, since 10 mm/s is not sufficient for PBT, it will not be used in rehabilitation applications only used to test the characteristics of the motor.

At a smaller position, rise and settling time do not significantly differ between velocities; however, as the position target increases, lower velocities' rise and settling time increase. The position tracking results are presented in Figures 28 and 29. As previous tables show the same result, as velocities increase, rise, and settling time

Table 7: Rise Time, Settling Time, and RMSE for different velocities at 10 mm position target

Velocities	Rise Time [s]	Settling Time [s]	RMSE
Velocity 10 mm/s	1.032 ± 0	2.328 ± 0.107	$4.840e-7 \pm 6.469e-8$
Velocity 20 mm/s	0.645 ± 0.001	0.955 ± 0.008	$5.255e-7 \pm 2.848e-7$
Velocity 30 mm/s	0.626 ± 0.090	0.955 ± 0.162	$4.287e-7 \pm 8.991e-8$
Velocity 40 mm/s	0.581 ± 0.011	0.912 ± 0.059	$3.356e-7 \pm 7.866e-8$
Velocity 50 mm/s	0.583 ± 0.003	0.932 ± 0.088	$3.052e-7 \pm 7.632e-8$
Velocity 60 mm/s	0.586 ± 0	0.972 ± 0.064	$3.436e-7 \pm 1.794e-7$
Velocity 70 mm/s	0.572 ± 0.004	1.552 ± 1.229	$2.792e-7 \pm 1.590e-7$
Velocity 80 mm/s	0.574 ± 0.007	0.996 ± 0.109	$3.328e-7 \pm 1.330e-7$
Velocity 90 mm/s	0.578 ± 0.006	1.056 ± 0.324	$2.272e-4 \pm 4.971e-4$
Velocity 100 mm/s	0.586 ± 0	0.971 ± 0.099	$2.463e-4 \pm 5.502e-4$

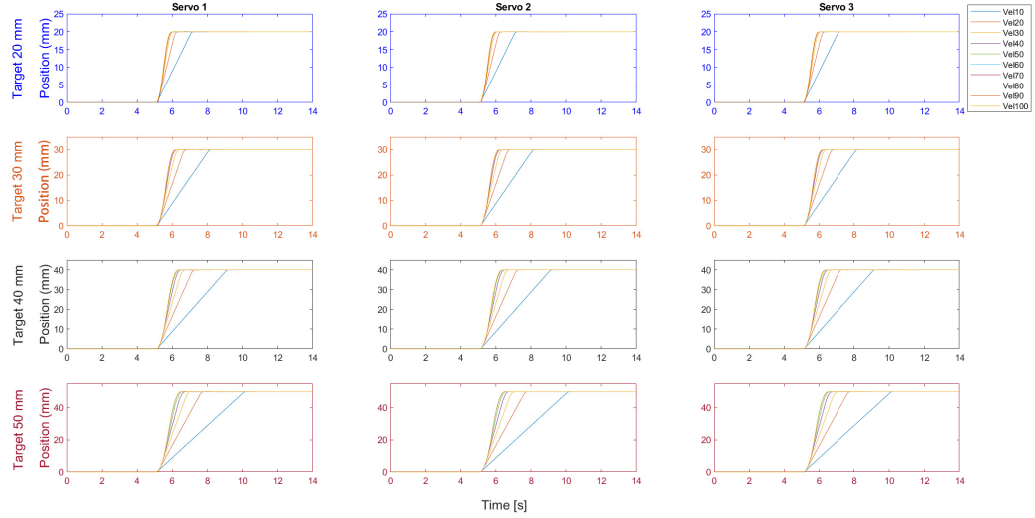


Figure 28: Servo motors step response for different velocities and position target between 20-50

decreases, these differences become more visible at higher positions.

5.3 Design Evaluation

The robotic platform design is evaluated by a series of experiments with a healthy subject; five repetitions of each experiment trial for 3 minutes with a 2 ms sampling rate are realized while the subject's feet are placed on the prototype. The testing is performed while the system is active and the user is passive (I-BaR framework first

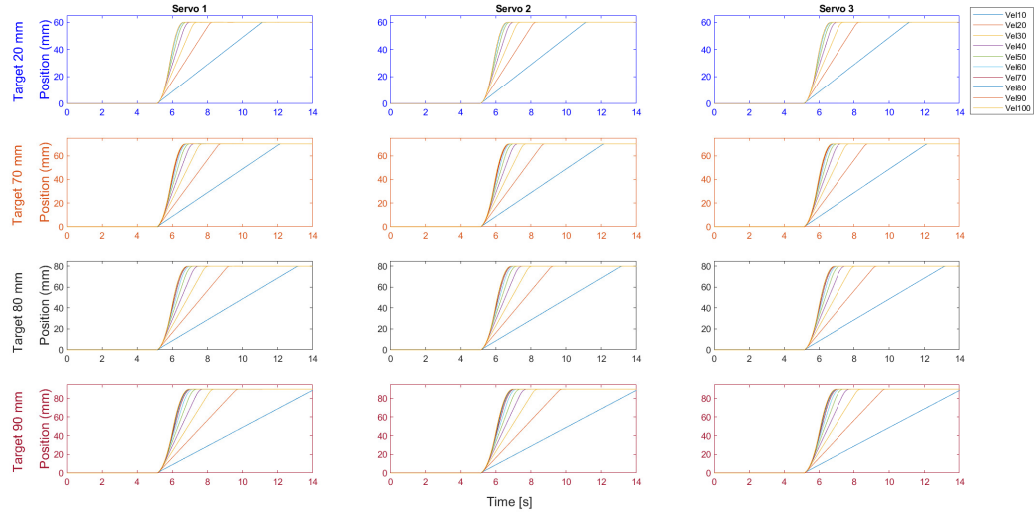


Figure 29: Servo motors step response for different velocities and position target between 60-100

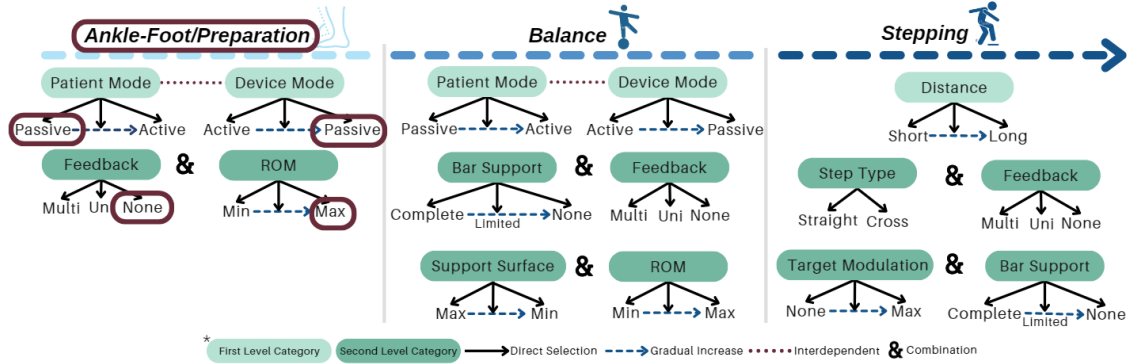


Figure 30: Design evaluation method selection based on I-BaR framework ankle preparation phase (Patient Mode: Passive, Device Mode: Active, Feedback: None, and ROM: Max)

testing (see Figure 30), i.e., ankle preparation phase (Patient Mode: Passive, Device Mode: Active, Feedback: None, and ROM: Max)).

The studies involving human participants are reviewed and approved by the University Research Ethics Council of Istanbul Medipol University under the E-66291034-772.02-4494 number. The participant provided their written informed consent to participate in this study.

These tests aimed to evaluate the system’s effectiveness in facilitating ankle preparation rehabilitation and its sensor communication between the electric and load cell panels. Each trial involved the subject performing specific exercise sets (subject passively placed her feet in the system’s maximum DoF limits), i.e., end effector rotation in roll and pitch and translation in z movements, then with the inverse kinematic equations (explained in kinematics section), corresponding servo motors’ positions are found and used as the reference trajectory.

Understanding the dynamic interaction between the servo motors and load cell sensors is crucial for ensuring the manipulator’s accurate control and reliable operation (between the load cell and electrical panels). Thus, the load cells, servo motor position tracking, and end effector orientation data throughout the rehabilitation process are analyzed during each reference orientation (roll, pitch, and z limits). The servo motors are effectively manipulated to attain their intended positions by employing specific end effector references (roll, pitch, and z) through the utilization of inverse kinematic equations. Subsequently, the measured end effector’s value is obtained from forward kinematic equations with linear actuators encoder data. The comparison between the values derived from these forward kinematic equations and the reference values is the first 10-second values represented in Figure 31. While the first two columns show the end effector orientation and position of the load cell during roll movement, the two in the middle show during pitch motion, and the last two show the movement in the z-translation. This comparison shows the alignment between the referenced orientation and the measured results. Servo Load cells experimental trials’ mean and standard deviation have been presented in Table 8, 9, and 10. The first 15-second values of the data collected for each load cell are presented in the Figure 32. While the first two columns show the measurement of load cell during roll movement, the two in the middle show during pitch motion, and the last two show the movement in the z-translation. Load cell numbers are divided according

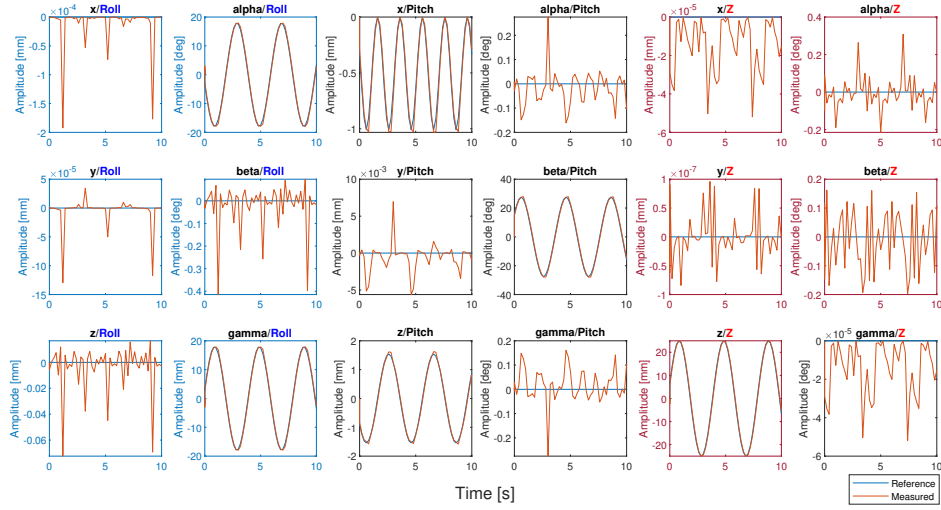


Figure 31: The evaluation of end effector configuration (x, y, z, alpha, beta, gamma) during roll, pitch, and z-translation movement, respectively (Blue (first two columns), black (third and fourth column), and red color (last two columns) text represents roll, pitch, and z-translation movement, respectively)

to the 4 sections on the footplate (see Figure 2I) starting from the left foot, i.e., the foot sections of the left foot are numbered from 1 to 4 and the right foot sections are numbered from 5 to 8. The distribution in the load cell weight reading is similar to the pressure transfer on the sole of the foot reported in the literature. As seen in the Table 8, 9, 10, approximately 50% of the weight goes to the sole, while the remaining 50% is distributed to the toes and metatarsals [173, 179]. The mean and standard deviation of each load cell showed approximately the same oscillations across each experimental trial. The multiple-trial sets proved the consistency and reliability of the system's performance across different sessions. Similarly, the servo motors position tracking for the first experimental trial is presented in Figure 33.

Furthermore, as seen in Figure 32, at each trial, load cell's measurements exhibit approximately the same oscillations across each experimental trial. This consistency in the oscillation patterns demonstrates the system's ability to consistently and accurately capture the applied forces during the exercises since they are conducted with

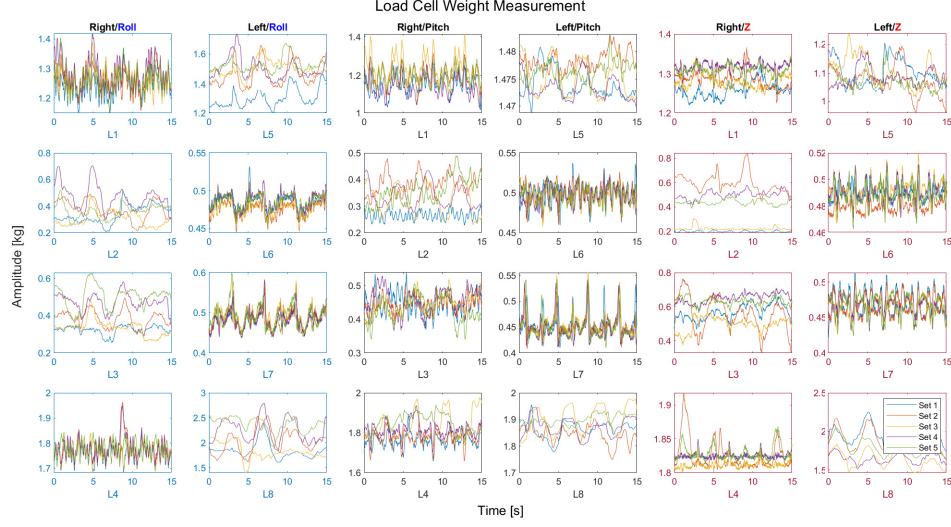


Figure 32: The evaluation of load cells during roll, pitch, and z-translation movement, respectively (Blue (first two columns), black (third and fourth column), and red color (last two columns) text represents roll, pitch, and z-translation movement, respectively)

the same healthy subject. It confirms that the system's sensors are capable of providing consistent data and making it a reliable tool for monitoring and assessing the force exerted during rehabilitation exercises. However, it is important to acknowledge some limitations that arise during this test. First of all, the evaluation is conducted with only one healthy subject. While this provides initial testing feedback, it may not fully represent the wide range of potential users with different foot sizes; thus, in the future, tests will be conducted with a larger population. Despite these limitations, the functionality tests demonstrate promising results, indicating that the system has the potential to be a valuable tool in the rehabilitation process for force measurement.

Understanding the dynamic interaction between the servo motors and load cell sensors is crucial for ensuring the manipulator's accurate control and reliable operation (between the load cell and electrical panels). Thus, the load cells, servo motor position tracking, and end effector orientation data throughout the rehabilitation process are analyzed during each reference orientation (roll, pitch, and z limits). The

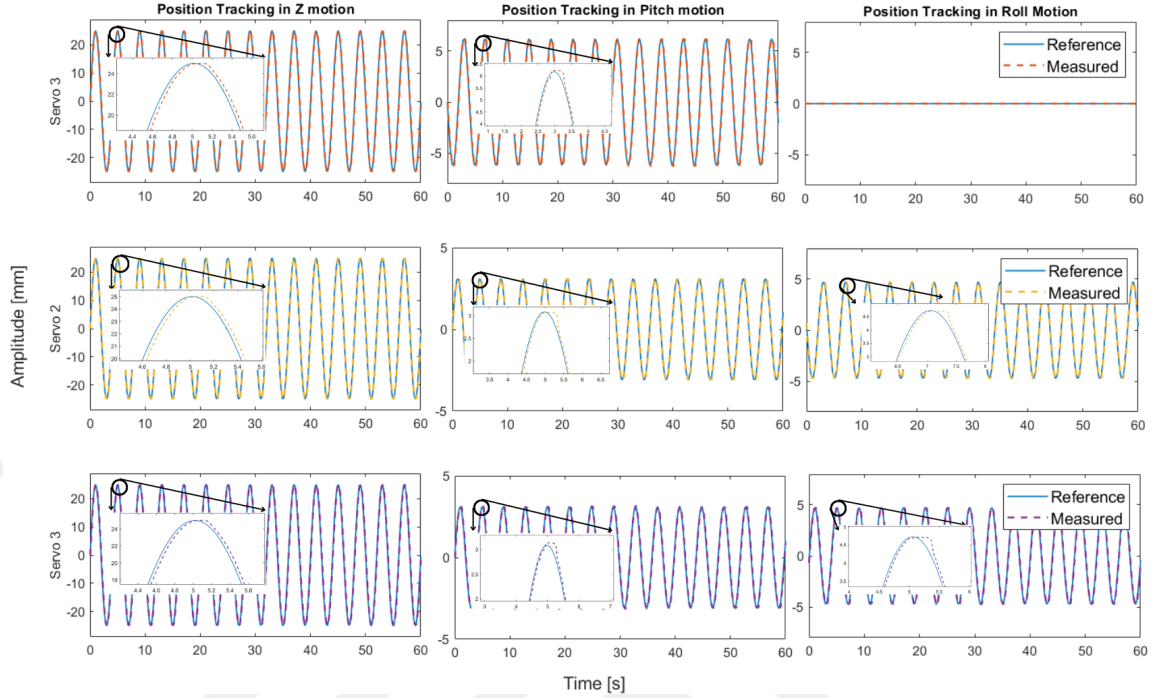


Figure 33: Response of each servo motor to the z-translation, and roll and pitch rotation

servo motors are effectively manipulated to attain their intended positions by employing specific end effector references (roll, pitch, and z) through the utilization of inverse kinematic equations. Subsequently, employing forward kinematic equations facilitated the found real orientation of the end function. The comparison between the values derived from these forward kinematic equations and the actual obtained values is the first 10-second values represented in Figure 31. While the first two columns show the end effector orientation and position of the load cell during roll movement, the two in the middle show during pitch motion, and the last two show the movement in the z-translation. This comparison shows the alignment between the referenced orientation and the measured results.

Table 8: Pressure distribution of the subject's feet while the end-effector is in z-translation motion

Load cell number	SET 1	SET 2	SET 3	SET 4	SET 5
LC 1	1.256±0.043	1.271±0.038	1.314±0.043	1.317±0.038	1.328±0.038
LC 2	0.328±0.189	0.514±0.076	0.490±0.151	0.493±0.060	0.521±0.064
LC 3	0.547±0.104	0.467±0.109	0.661±0.100	0.661±0.061	0.677±0.058
LC 4	0.526±0.060	0.521±0.024	0.520±0.020	0.526±0.024	0.527±0.022
LC 5	1.058±0.064	1.051±0.063	1.101±0.044	1.112±0.067	1.125±0.060
LC 6	0.486±0.015	0.479±0.015	0.493±0.043	0.490±0.045	0.488±0.049
LC 7	0.469±0.050	0.470±0.051	0.463±0.043	0.465±0.045	0.467±0.049
LC 8	1.846±0.148	1.841±0.127	1.711±0.098	1.828±0.137	1.765±0.086

Table 9: Pressure distribution of the subject's feet while the end-effector is in roll rotation motion

Load cell number	SET 1	SET 2	SET 3	SET 4	SET 5
LC 1	1.247±0.071	1.268±0.070	1.270±0.075	1.284±0.077	1.252±0.035
LC 2	0.383±0.089	0.356±0.055	0.386±0.115	0.418±0.109	0.321±0.103
LC 3	0.358±0.053	0.413±0.047	0.404±0.068	0.460±0.059	0.375±0.086
LC 4	1.799±0.145	1.774±0.069	1.770±0.066	1.805±0.106	1.747±0.094
LC 5	1.405±0.101	1.457±0.073	1.498±0.084	1.566±0.104	1.526±0.062
LC 6	0.475±0.026	0.476±0.025	0.481±0.024	0.486±0.024	0.484±0.024
LC 7	0.491±0.073	0.486±0.067	0.488±0.072	0.508±0.088	0.490±0.074
LC 8	1.890±0.231	1.126±0.193	1.924±0.315	2.332±0.283	2.332±0.226

Table 10: Pressure distribution of the subject's feet while the end-effector is in pitch rotation motion

Load cell number	SET 1	SET 2	SET 3	SET 4	SET 5
LC 1	1.174±0.080	1.209±0.082	1.211±0.082	1.243±0.090	1.253±0.098
LC 2	0.477±0.139	0.584±0.135	0.582±0.157	0.604±0.218	0.533±0.188
LC 3	0.589±0.103	0.729±0.094	0.873±0.133	0.956±0.114	0.847±0.121
LC 4	1.862±0.159	1.794±0.102	2.057±0.215	1.980±0.249	1.891±0.158
LC 5	1.215±0.057	1.280±0.087	1.259±0.094	1.304±0.104	1.320±0.146
LC 6	0.507±0.028	0.498±0.027	0.493±0.025	0.496±0.025	0.493±0.025
LC 7	0.457±0.051	0.456±0.052	0.455±0.052	0.447±0.046	0.448 ±0.049
LC 8	1.888±0.077	1.834±0.145	2.097±0.159	1.937±0.155	1.979±0.193

5.4 *Discussion*

This study investigates the communication and performance characteristics of the servo motors' and load cell sensors to improve overall system performance and capabilities during further prototype construction and rehabilitation testing. Two panels are created to enable the system to communicate, namely a load cell panel and an electrical panel. The reason for the separate panels is to keep load cells' analog signal as long distances as possible so that the analog signal is not affected by 50 Hz power line noise. One of the design testing aims is to analyze the communication of these two panels.

The characteristics of the servo motors and the design of the system are evaluated to see the 3 DoF parallel manipulator and panels' behavior during the ankle preparation phase. In the context of the characterization, a chirp signal (initial frequency starting at 10 Hz and ending at 1 kHz for 120 seconds) is given as a reference input to the assembled platform, and the output signal is recorded across various frequencies. The crucial facet of the analysis involves the identification of the cut-off frequencies. This frequency indicates a decline in the motor's capacity to precisely and consistently react to input signals. It is vital to stay within the frequency range because, in human-subject experiments, the system must be stable. As the system operates with all motors within a unified control loop, the cut-off frequency for the entire setup is dictated by the lowest frequency among the motors, which is determined to be 6300 Hz (see Figure 26). The end effector is not symmetrical, so during characterization, a higher force is applied to the first motor which is why its cut-off frequency is found to be lower. This cutoff frequency holds significant importance not only in terms of stability but also in determining the velocity of perturbation. Considering the I-BaR velocities limitation, the results derived from the characterization tests are found to be satisfactory.

The second servo motor characteristics test is testing the motors on different positions (10-100 mm) and velocities (10-100 mm/s) with accuracy, rise and settling time. The maximum RMSE values are obtained as 0.773 mm when servos are given 100 mm position and 100 mm/s velocity at first servo motors. This position error results in a maximum error of 0.24 degrees in the orientation of the end effector. Although this angle value does not make a significant difference for healthy people, it may be an important value in patients with high spasticity. While this characteristics test provides valuable insights into the performance of the motors, it also revealed certain limitations that should be acknowledged during ankle preparation rehabilitation. This error is due to the non-linear kinematic equations and mechanical imperfections. Such problems are tackled by utilizing CTC to alleviate non-linearity and replacing the ABS material used for the joints with aluminum, respectively. Moreover, the experiment identifies that the maximum rise and settling times occurred when the motors are operated at a position of 100 mm and a relatively lower velocity of 10 mm/s. Additionally, the observed higher torque requirements time during this motion can cause mechanical stress on the system, which could impact its overall longevity and reliability.

The design functionality of the system is tested with one healthy subject with five experimental trials with a three minutes duration and 2 ms sampling rate. Approximately 50% of the weight distribution went to the sole, while the remaining 50% is distributed to the toes and metatarsals as stated in literature. Furthermore, as seen in Figure 32 at each trial, the load cell's measurements exhibit approximately the same oscillations across each experimental trial. This consistency in the oscillation patterns demonstrates the system's ability to consistently and accurately capture the applied forces during the exercises since they are conducted with the same healthy subject. It confirms that the system's sensors can provide reliable and consistent data

and make it a reliable tool for monitoring and assessing the force exerted during rehabilitation exercises. However, it is important to acknowledge some limitations that arise during this test. First of all, the evaluation is conducted with only one healthy subject. While this provides initial testing feedback, it may not fully represent the wide range of potential users; thus, tests will be conducted with a larger population in the future. Despite these limitations, the functionality tests demonstrate promising results, indicating that the system has the potential to be a valuable tool in the rehabilitation process for force measurement.

The maximum mean RMSE of position tracking is 0.699 mm at z translation. These position errors result in 0.24 degrees of error in the orientation of the end effector. When the corresponding distances are been analyzed, these values are higher than the Simscape CTC simulation. The error arises from distinctions in physical and simulation environments, such as joint friction and sampling intervals (simulation involves continuity, whereas real-time operation involves signal discretization). Moreover, mechanical imperfections and the non-linearity in kinematic and dynamic equations also contribute to this variation. Consequently, our future aim also involves the application of diverse control techniques aimed at mitigating tracking discrepancies.

Overall, the impact of these characteristics and design tests is evident in the system's ability to track movements and collect data successfully during the ankle preparation phase. The results provide valuable insights into the designed system's performance, serving as a foundation for further refinement and improvement of the system's design. In addition, in this article, the system design is tested on ankle preparation, in other words, the load cell evaluation was made while the system was active and the user was passive (I-BaR framework first testing (Patient Mode: Passive, Device Mode: Active, Feedback: None, and ROM: Max)). In future studies, different control methods will be tested on the servo motor in which patients can actively

participate for further I-BaR framework testing (impedance control and CTC, etc.). In addition, patients to be included in the study should be able to stand for at least 60 seconds and walk at least 10 m without manual assistance. Furthermore, their Functional Ambulation Scale (FAS)<4, Berg Balance Scale (BBS)<40, and Modified Ashworth Scale (MAS)<3 levels should be in the specified range. Thus, the system will be improved to be compatible with the above-mentioned patient group. Moreover, the remaining parts of the system will be prototyped, and balance and stepping rehabilitation will be studied.

CHAPTER VI

CONCLUSIONS

The proposed novel BalanSENS system allows three different uses in accordance with the I-BaR method. In the first phase, the patient is seated in the chair while the ankle ROM, strength, proprioception, and sensory training are performed by the 3-DoF parallel manipulator. The patient's active participation level can be adjusted, and the contribution to motor learning can be increased (see Figure 1A). The perturbation given by the parallel manipulator starts with the angle value explicitly determined for the patient, and it is aimed to reach the ankle ROM limits of healthy individuals at the end of the phase. In the second phase (see Figure 1B), PBT is implemented with the 3-DoF parallel manipulator. Moreover, haptic bars are designed to measure how much help the patients need from their upper extremities to control their balance during PBT. These bars contain load cells and vibration motors, so required support can be quantitatively measured, and haptic feedback can be given when the support threshold is exceeded. In addition, the angle between the footplates (foot rotation angle) is adjustable between $0-20^\circ$, so the difficulty of treatment can be adjusted by decreasing/increasing the support surface during rehabilitation. In the last phase (see Figure 1C), patients are asked to take steps depending on the external signal to the target points that are placed straight and diagonally. In this way, it is possible to investigate whether the changes in the initiation and reaction time of step-taking are affected by the task distance. GRF can be measured and haptic feedback can be given from the sole on target points since each target includes footplate. The linear rail system can modulate the distance between these target points, and perturbation can be provided on the x-y plane. Moreover, patients can take help from the haptic

bar while stepping, as in the balance rehabilitation threshold, which is used to gain independence gradually. In addition, the movement of the patient's upper extremity and trunk can be observed with the IMU ring placed on the patient during both the balance and stepping phases. Thus, BalanSENS eliminates the shortcomings of the systems mentioned in the literature and can provide quantitative measurement.

The system design is appraised by the simulations to detect the required specifications for the servo motors, and evaluate the load-bearing capacity and feasibility and the CTC method. In the light of simulation outcomes, the physical robotic platform is prototyped. The servo motor characteristics are evaluated through the application of chirp signals, revealing their adequacy for the envisioned application. Moreover, the examinations involving a healthy volunteer carried out to prove the system design functionality exhibited encouraging outcomes in force measurement and real-time data monitoring during rehabilitation routines, despite some constraints. The findings of this study provide a groundwork for our subsequent improvements. Future studies will include stroke patients with mild-to-moderate motor impairments in the affected ankle with foot gait abnormality in the ankle phase. After the ankle phase, the balance phase will be started, and trunk sway, CoM deviation, reaction time, and force distribution in the haptic bar will be evaluated in the patients. Final rehabilitation tests include the stepping phase and the reaction time and weight transfer capacity will be evaluated.

The system design is appraised by the simulations to detect the required specifications for the servo motors and evaluate the load-bearing capacity and feasibility. The physical 3 DoF platform is implemented in light of simulation outcomes to facilitate communication between servo motors and sensors. The servo motor characteristics are evaluated through the application of chirp signals, revealing their adequacy for the envisioned application. Moreover, the examinations involving a healthy volunteer carried out to prove the system design functionality exhibited encouraging outcomes

in force measurement and real-time data monitoring during rehabilitation. Despite some limitations, the findings of this study provide a groundwork for I-BaR Framework.



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