Aalto University

School of Electrical Engineering

Bachelor of Electrical Engineering

Design Considerations for Lower-Limb Exoskeletons

Bachelor's thesis

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TEKNIIKAN KANDIDAATINTYÖN TIIVISTELMÄ

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Liikkumiskyky on keskeinen toiminto, joka parantaa yksilön itsenäistä toimintakykyä ja hyvinvointia. Neurologiset sairaudet, vammat ja ikääntyminen voivat kuitenkin johtaa kävelyhäiröihin, jotka heikentävät elämänlaatua ja kasvattavat avuntarvetta. Kuntouttavat ja avustavat alaraaja-eksoskeletonit, eli ulkoiset robottiset tukirangat, tarjoavat uudenlaisia ratkaisuja näihin haasteisiin. Ne tukevat ja tehostavat kävelyä, ja niitä voidaan käyttää henkilöillä, joilla on esimerkiksi aivohalvaus, selkäydinvamma, Parkinsonin tauti tai ikääntymiseen liittyviä motorisia ongelmia. Väestön ikääntyessä ja kuntoutus- sekä avuntarpeiden kasvaessa eksoskeletonien merkitys korostuu entisestään. Nykyiset laitteet kuitenkin kohtaavat haasteita laitteiden omaksumisessa, mitkä liittyvät suunnittelun ja käyttäjien tarpeiden yhteensovittamiseen.

Tämän kandidaatintyön tavoitteena on tarkastella kuntouttavien ja avustavien alaraaja-eksoskeletonien keskeisiä suunnitteluominaisuuksia ja niiden yhteyttä kuntoutustavoitteisiin sekä käyttäjälähtöisiin tarpeisiin. Työ toteutettiin kirjallisuustutkielmana olemassa olevien alaraaja-eksoskeleton -tutkimusten pohjalta. Tämä näkökulma mahdollistaa analyysin teknisestä näkökulmasta ja käyttäjien tarpeista. Tutkimus on rajattu lääketieteellisiin, avustaviin ja kuntouttaviin eksoskeletoneihin keskittyen ihmisen ja eksoskeletonin väliseen vuorovaikutukseen kliinisessä ja päivittäisessä käytössä. Teollisuus- ja sotilaskäyttöiset laitteet on jätetty tutkimuksen ulkopuolelle. Lisäksi työ ottaa huomioon erityisesti kotikäytön ja arjen ympäristöjen vaatimukset, joissa laitteen helppokäyttöisyys ja mukautuvuus korostuvat.

Tämä tutkielma esittelee eksoskeletonien suunnittelua kolmesta näkökulmasta: askelluksen biomekaniikasta, eksoskeletonien mekaanisista ominaisuuksista ja käyttäjälähtöisestä suunnittelusta. Aluksi käsitellään standardia askellusmallia ja sen häiriöitä eri sairauksien seurauksena. Askelhäiriöiden ymmärtäminen on tärkeää, koska yksilöiden biomekaaniset tarpeet toimivat pohjana teknisten ratkaisujen ymmärtämiselle. Toisena analysoidaan eksoskeletonien mekatroniikkaratkaisuja, kuten nivelten moottoreita, liikeratoja ja materiaalivalintoja, sekä painon ja voiman tasapainoa laitteen siirrettävyyden kannalta. Käyttäjien näkökulmasta korostetaan laitteen mukavuutta, turvallisuutta ja helppokäyttöisyyttä laitteen hyväksyttävyydelle käyttäjien näkökulmasta. Käyttäjälähtöinen suunnittelu mahdollistaa yksilöllisen säätämisen eri käyttäjien tarpeisiin, mikä on erityisen tärkeää monipuolisessa potilasryhmässä.

Johtopäätöksenä tutkimus korostaa tarvetta yhdistää tekninen suunnittelu ja käyttäjien monipuoliset tarpeet, jotta eksoskeletoneista voidaan kehittää tehokkaita, turvallisia ja saavutettavia apuvälineitä kuntoutukseen ja avustukseen. Tasapainoittelu teknisten ominaisuuksien ja käyttäjien tarpeiden välillä on haastavaa, mikä vaikuttaa eksoskeletonien suunnittelua. Esimerkiksi laitteen keventäminen parantaa käyttäjän mukavuutta, mutta samalla se voi heikentää laitteen tarjoamaa avustavaa voimaa, mikä saattaa rajoittaa sen soveltuvuutta käyttäjän tarpeisiin. Tulevaisuuden kehityksessä painopisteen tulee olla käytettävyyden ja mukavuuden parantamisessa sekä kustannusten alentamisessa, jotta laitteista saadaan hyödyllisiä työkaluja liikkumiskyvyn tukemiseen.

Avainsanat: alaraaja-eksoskeleton, käyttäjälähtöinen suunnittelu, ihmisen biomekaniikka, toimilaitteet ja ohjaus

AALTO UNIVERSITY BACHELOR’S THESIS

SCHOOL OF ELECTRICAL ENGINEERING ABSTRACT

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Walking is one of the key abilities of humans. As the population is aging and the number of neurological impairments is increasing, walking dysfunctions are becoming more frequent. Lower-limb exoskeletons serve as a possible solution to rehabilitate this population. These devices help restore mobility and enhance rehabilitation in people with walking impairments. This bachelor’s thesis reviews current design choices from three different perspectives, human biomechanics, engineering solutions, and user-centred-design, to examine how these align together in clinical and home use settings with the goal of highlighting future directions of development. The objective of the thesis is to examine how technical aspects of exoskeletons and user needs meet. The research is limited to rehabilitative and assistive exoskeletons.

The thesis is conducted as a comprehensive literature review and explores lower-limb biomechanics and common gait pathologies highlighting how these conditions influence exoskeleton requirements and design. Technical aspects are analysed, with a focus on efficient, wearable, and safe designs. Particular attention is given to actuator compliance, sensor integration, and adaptive control, which are essential for enabling natural movement. User-centred design considerations are explored, including the most important user requirements.

The study concludes that exoskeleton design must carefully balance multiple competing requirements, as enhancing one feature often affects others. Findings show that many current devices are too heavy, expensive, and complex for independent home use. User expectations and rehabilitation goals vary, which further emphasises exoskeleton design considerations. Achieving an optimal balance is essential to create comfortable and user-friendly devices, aligned with individual needs, making users more likely to adopt and continue using them. Improvements in current technology are essential to satisfy user requirements.

Keywords: lower-limb exoskeleton, user-centred-design (UCD), gait rehabilitation, human biomechanics, actuators and control

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1 Introduction

Exoskeletons are wearable robotic devices, which are designed to support and improve human movement. They provide increased individual capabilities and are used as aid when performing certain tasks [1]. The key purposes of exoskeletons include rehabilitation, mobility assistance, and physical therapy and training. Exoskeletons assist patients with neurological and musculoskeletal disorders, such as stroke and spinal cord injury, in regaining mobility. They support individuals with limited or impaired movement.

According to Global Burden of Disease study [2], 2.4 billion people worldwide are affected by disorders, with an estimated 255 million suffering from neurological disorders. This figure does not even include the number of lower- and upper-limb disfunctions, such as stroke-related impairments, which affect approximately 86 million people. In addition, some of the musculoskeletal disorders, including lower-limb amputation, osteoarthritis and other injuries impacting lower-limb function, should be considered within the target group. Although the exact number of individuals with lower-limb impairments is unknown, the overall amount of people is significant, highlighting the large number of people impacted by lower-limb disorders.

Spinal Cord Injury (SCI) and stroke are the most common neurological disorders causing impairment. There is an estimate of 12 500 new SCI cases annually in North America [3]. Most countries manifest stroke as the second or third most common causes of death, e.g. in the USA alone, close to 800 000 people experience stroke every year [3]. The World Health Organization (WHO) reports 15 million global stroke diagnoses annually, roughly 6 million of whom lose their lives and 5 million are develop permanent impairments [4]. Long-term disabilities often affect the elderly, who are also associated with the highest amount of stroke incidences [5]. Among those who have suffered from a stroke, 26% are left with disabilities in basic daily activities and 50% suffer from hemiparesis-caused decreased mobility [5]. Approximately 60% of patients with neuromuscular disorders suffer from gait disorders which highly impacts their quality of life [6].

Elderly people are also becoming an increasingly important target group as mobility functions decrease with aging, portraying that assistive devices might soon be in high demand to support daily lives. By 2029, people above the age of 65 will account for over 20% of the entire US population, resulting in an increased demand for preventive, acute, and rehabilitative health-care services and tools to enable independent functioning in daily activities [7] p. 5. At present, approximately 11.5% of the entire United Nations population are at least 60 years old and this number is predicted to almost double by 2050 [8]. Consequently, exoskeleton devices are needed to benefit impaired and elderly people [8].

Exoskeleton rehabilitation can assist walking for impaired people, which is crucial for a human's internal systems and organs' correct functioning. Lower-limb movement is essential for a functional body, offering many health benefits: stabilised blood pressure, improved pulmonary ventilation, prevention of the degeneration of muscle and bone tissue, and increased joint mobility [1]. Additionally, other health benefits were identified, such as relieved pressure, improved circulation, enhanced bone density, improved bowel and bladder function, reduced potential for developing orthostatic hypotension, and general advantages related to standing and walking, such as improved skin integrity, reduced spasticity, and pain relief [9]. Given these significant health benefits, the early initiation of exoskeleton rehabilitation is crucial.

Research suggests that early and intensive rehabilitation is more effective in promoting neuroplasticity [10]. It is important to not only activate the prefrontal cortex but also the brain extensively, as the primary somatosensory cortex, premotor area, primary motor cortex, and supplementary motor area all play essential roles in motor learning [11]. The patient's undamaged brain area is activated by the motion which restores the neural network of the brain [7]. Neuroplasticity plays a critical role in rehabilitation following neurological injury or disease [10]. Neuroplasticity is a process that allows the brain to adapt to changes, recover lost function, and improve cognitive and motor skills [10]. Neurorehabilitation, a neuroscience-based approach, aims to repair neural circuits by enforcing repetitive walking movements of the legs. This method can be used for stroke and SCI patients to restore a more natural gait [11].

Despite the proven benefits and an increasing demand, exoskeleton adoption outside academic research remains challenging, owing to a set of design issues. In this thesis, the main identified problems with exoskeletons are comfort, ease of use, and cost. Lower-limb exoskeletons are generally unsuitable for home use, which is a significant part of effective rehabilitation. It is important that such devices are safe, lightweight, and simple to don and doff [7]. For achieving comfortability, weight continues to be the main challenge. Issues with portability and heavy weight reduce user comfort and are due to the mechanical constraints which are presented in more detail in chapter 3. The survey of stakeholder perspectives stated that the cost of exoskeletons poses one of the largest concerns due to the unknown cost effectiveness of robotic devices [9]. Generally, exoskeletons are not individualised for different users. Individualization improves the user-centred design, such as comfort and ease-of-use of exoskeletons, since the user's characteristics and needs can be identified [7]. These issues are addressed more closely in Chapter 4.

This thesis examines the design choices of lower-limb exoskeletons, analysing their properties and functionalities in relation to the diverse demands of human mobility. The scope includes rehabilitative exoskeletons in clinical settings and assistive exoskeletons in everyday use. Exoskeletons used for industrial or military applications are excluded in this thesis. The research focuses on human-exoskeleton interaction, usability factors, and user-centred design principles, while the technical aspect of exoskeletons are examined in relation to their design and usability. This thesis is conducted as qualitative research, including a comprehensive literature review analysing existing studies on lower-limb exoskeletons, their technical aspects, and user requirements. The aim is to unify different perspectives on exoskeleton design, identify limitations, and highlight areas for future development.

2 Biomechanics of human lower-limb and gait pathologies

Walking is an intricate process that includes coordinated movements and forces enabling locomotion. It is essential to understand this process when designing walking assistive devices such as lower-limb exoskeletons [12]. This is due to mobility being significantly impacted by even small issues in gait. Different factors such as disorders or injuries can impact gait pathologies in various ways, requiring tailored assistance and rehabilitation. To fully understand how exoskeletons can support these needs and how their design choices can affect this support, this chapter explores the background of the biomechanics of human walking, provide a detailed description of standardized gait phases, and explore the various factors and disorders that contribute to abnormal gait, as well as their effects on walking patterns.

2.1 Standardised gait

Walking is commonly thought as a daily life activity, yet it is highly complex [13] and often unrecognised, since it is learned and practiced throughout a lifetime. Walking requires the use of every nervous system level with multiple elements of the musculoskeletal system among the cardiorespiratory system [13]. Gait refers to the characteristic pattern of human walking [7] pp. 171-172. As a core function, bipedal gait plays a significant role in shaping a human's life, being nearly as important as speech, advanced analytical abilities, and use of complicated tools [13]. For a standardised gait, each of the following functions are needed: locomotor function (for sustaining rhythmic gait), postural reflexes, motor control, sensory function and sensorimotor integration, balance, the musculoskeletal system, and cardiopulmonary functions [13]. Typical gait uses the limbs and trunk to combine rhythmic, systemic, and coordinated movements to collectively create forward movement [4]. A person's gait pattern is modified by factors such as age, personality, mood, and sociocultural influences [13]. There are significant challenges in trying to determine a normal and appropriate range for many gait features, since a wide variety of gait patterns between varying age groups and genders is clearly visible [4].

Walking consists of the gait cycles, which can be subdivided into seven gait phases [7]. Each phase begins with a specific gait event, resulting in a total of seven gait events. Gait phases and events highlight the sequential movements necessary for bipedal locomotion and are essential for analysing standardised and pathological gait patterns. The gait cycle consists of two phases, being the stance and swing phase, while optionally being further subdivided into seven gait phases: loading response, mid-stance, terminal stance, pre-swing, mid-swing, and terminal swing [7], [13]. Gait events and gait phases in one gait cycle are visualised in Figure 2-1. The stance phase continues for about 60% of the gait cycle, corresponding to the duration between heel strike and toe-off of the same foot [4], [13]. The stance phase begins and ends with both feet being on the ground [13]. Both periods account for 10-12% of the gait cycle individually, while providing double support [13]. The swing phase lasts for approximately 40% of the gait cycle [13], beginning with toe-off and ending with heel contact of the foot [4]. Each seven gait phases are denoted as gait events, which are initial contact, opposite toe off, heel rise, opposite initial contact, toe off, feet adjacent, and tibia vertical [7]. These gait events represent the transitions of the gait phases, which moves the body forward [7]. Gait events occur at certain stages in a specific order of the gait cycle during standardised walking [7].

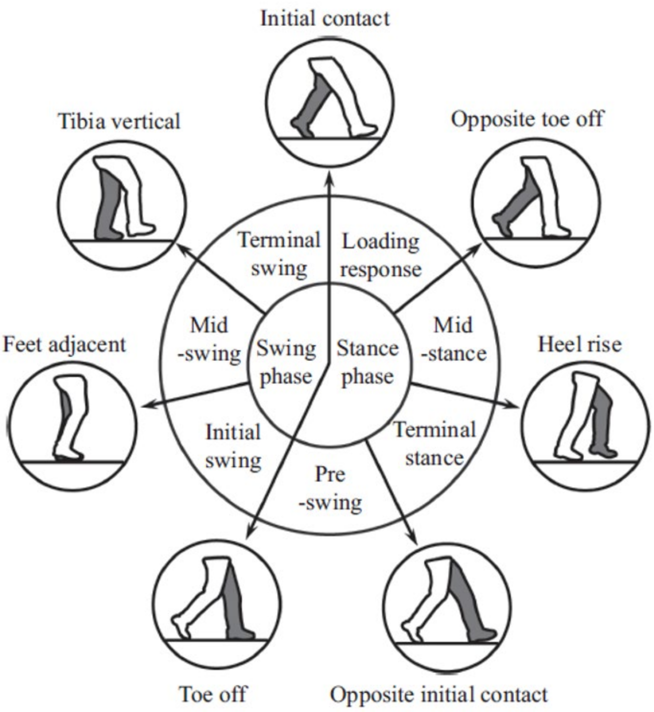

Figure 2-1. Gait events and gait phases in one gait cycle, adopted from [7] *p. 171.*

There are certain walking parameters that healthy adults should be able to reach. The preferred walking speed is approximately 1.4 m/s up to 59 years old adults [13]. The preferred walking speed declines approximately by 1% per year in healthy elderly subjects, leading to a mean of 0.95m/s in individuals over the age of 80 years [13]. According to [4], the step length range with healthy adults is 0.68-0.85 m. It has been reported [13] that young adults' average cadence ranges from 115 to 120 steps per minute. While cadence remains comparably stable with aging, gait speed and step length decline [13]. Also, the elderly often reach a wider base of gait to gain better balance and for a reduced fear of falling [13]. Walking speed, cadence, walking base width, step length, and stride length are central measurements of gait [13]. These gait measurements offer a standardised approach to access and quantify an individual's walking ability.

2.2 Human Walking Biomechanics: Joint Movements and Forces

Biomechanics explain how the muscles and joints generate the forces, which produce movement. In exoskeleton design, it is especially valuable to know the power and angle requirements for each joint [12]. Coordinated and adaptive movement patterns are formed when said interactions occur at each joint within defined degrees of freedom (DOFs). For walking, biomechanics is used to describe the movement and adaptation of the body during each step. This forms the foundation for gait analysis and exoskeleton design.

Human legs have a seven DOFs structure [8], [12], allowing complex and versatile movement, such as walking. The seven DOFs compose from the hip having three rotational DOFs, knee having one, and the ankle having three [8], [12]. The hip is a ball-and-socket joint, whose DOFs are abduction/adduction, flexion/extension and hip intra/extra rotation [8], [12]. The knee joint only has a single DOF in the sagittal plane, flexion/extension, and the ankle joint's three DOFs are abduction/adduction, dorsi/plantarflexion and inversion/eversion [8], [12]. Figure 2-2 illustrates the DOFs in the lower limbs, depicting the possible movement directions at each joint. The coordination and activation of these seven DOFs enable smooth locomotion by facilitating the necessary movements and adjustments at each step and gait phase.

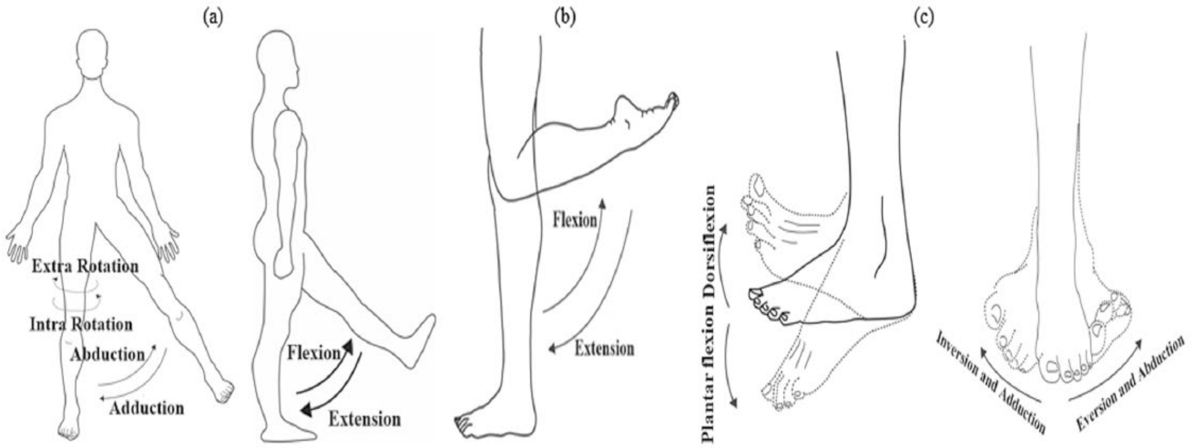

Figure 2-2. The seven degrees of freedom (DOFs) of lower-limb joint movements [8]. *a) shows the hip abduction/adduction, intra/extra rotation and flexion/extension; in the middle b) knee flexion/extension; and on the right, c) ankle dorsi/planter flexion, abduction/adduction, and ankle inversion/eversion. The figure is adopted from* [8].

All DOFs are essential to execute the gait cycle. By combining the information from Figures 2-1 and 2-2, it can be interpreted which DOFs are important in each phase of the gait cycle and during specific gait events. It can be observed that the hip flexion is important throughout the swing phase and the extension is important during terminal stance and pre-swing. Hip abduction and adduction are important for pelvic stability during mid-stance and the hip internal and external rotation are important for foot placement in terminal swing and loading response phases. Knee flexion is critical for loading response and initial swing phases and extension in mid-stance and terminal swing. Ankle's dorsiflexion is important for initial swing and mid-swing and plantar flexion for push-off, which consists of heel rise and toe off. Ankle abduction and adduction are important for foot placement and balance throughout the stance and swing phases. Lastly the ankle inversion is important for heel rise and terminal stance and the ankle eversion for loading response by supporting the ankle stability.

Understanding the biomechanics of walking requires analysing the movement patterns, torques, and forces that occur at the lower-limb joints throughout the gait cycle. Figure 2-3 represents joint angles, joint moments, and joint power for the hip, knee, and ankle joints throughout a full gait cycle (0-100%, starting and ending at heel strike). The first frame of Figure 2-3 represents the changes in joint flexion and extension angles in radians. The positive values refer to the flexion of the joint and the negative values to the extension of the joint, e.g. knee extension during the late stance phase. For example, the knee angle grows intensively for approximately 60% of the gait cycle, which refers to the knee flexion in the swing phase.

The middle frame represents the joint moments/torques, caused by muscle forces and external forces. Positive torque drives the muscle to produce flexion, and correspondingly negative torque drives the muscle to extension. For instance, the knee joint's moment becomes negative after the opposite initial contact/heel strike, meaning that the quadriceps muscles must generate an extension torque to keep the knee joint stable during the stance phase. Also, the hip joint generates a large moment simultaneously at heel strike, while the hip flexor develops torque to lift the leg during swing. The ankle joint moment curve shows a significant drop into negative values, representing the plantarflexion moment during late stance. In this phase, the ankle generates torque to push off the ground, enabling forward motion.

The last frame shows the power for each joint. The power is produced by the joint moment and angular velocity. In joint power, the positive value means that the muscle is producing energy, e.g. the ankle's push-off. Negative power on the contrary means that the muscle is absorbing energy, e.g. the knee slowing down after heel strike. The greatest power peak appears in the ankle joint approximately 50-60% into the gait cycle, responding to the push-off phase, where the ankle plantar flexion (illustrated in *Figure 2-2*) generates forward propulsion in the pre-swing phase.

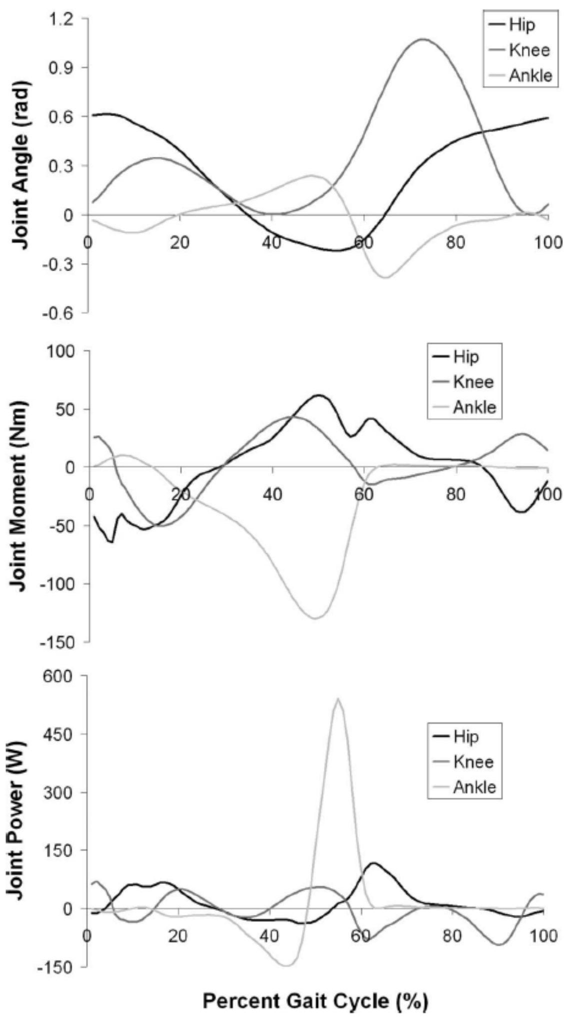

Figure 2-3. Represent the angles, moments and power of the leg flexion/extension joints over the gait cycle while walking at 1.27 m/s speed. The cycle begins and ends to the heel strike. The figure is adapted from [12] *p. 3.*

2.3 Factors contributing to abnormal gait

All individuals' natural walking style is different [14]. However, medical conditions and injuries can affect the walking pattern and contribute to abnormal gait [14]. Gait disorders contribute to impaired quality of life by decreasing personal freedom and increasing the risk of falls and injuries [13]. Gait disorders result from various causes, including neurological conditions, e.g. motor or sensory impairments, orthopaedic problems, e.g. osteoarthritis and skeletal deformities, and other medical conditions [13]. Gait disorders reduce the DOFs in pathological conditions, as many of them effectively reduce, or in some cases, prevent the active range of motion in certain joints.

Abnormal gait can be caused by anything that affects the brain, spinal cord, legs, or feet [14]. Some frequent examples of abnormal gait include limping, dragging toes, shuffling feet, short steps, difficulty of supporting one's own weight, and trouble coordinating [14]. Gait abnormalities include various different types of gait patterns, the most common being: Antalgic gait (avoidance of pressure on the affected leg while stepping), Propulsive gait (Parkinsonian gait, includes stooping, rigid posture and bending forward), scissors gait (hitting of knees and thighs or crossing in a scissors-like pattern), spastic gait (hemiplegic gait, causes one to walk with a stiff leg), steppage gait (neuropathic gait, causes a high step, where the hip is elevated to lift the leg higher than normal), waddling gait (causes overstated movement of the upper body, which leads to a waddling walk), and crouching gait (causes the ankles, knees and hips to flex while walking) [14]. The most common gait abnormalities causing health conditions include stroke, multiple sclerosis, arthritis, hemiplegia, cerebral palsy, Parkinson's disease, and spinal stenosis [14]. This section will focus on the following common causes of gait disorders: stroke, spinal cord injury (SCI), Parkinson's disease, and aging.

2.3.1 Stroke

Stroke is a leading cause of adult disability, leaving a permanent gait dysfunction for over two thirds of stroke survivors, which negatively affects functional mobility and quality of life [15]. Post-stroke gait impairments are typically described by abnormal kinematic (joint angle motion) and kinetic (forces, moments) patterns, altered muscle activation, deviations in the spatiotemporal features, and increased energy consumption during walking [4], [15]. Stroke patients may experience deficiencies in mobility, muscle strength and tone, perception and motor control, sensation, and balance, leading to possible involuntary movements that affect gait patterns [4]. Kinematic impairments caused by stroke in the above-mentioned muscle strength, motor function, and balance are strongly connected to the ability of walking [4].

Stroke-caused kinematic deviations in gait appear in various ways in different gait phases. In [16], it is illustrated how stroke-related changes affect joint motion during walking. At the initial stance, gait deviations are limited to ankle dorsiflexion and lack of knee flexion [16], which decreases the anterior tibial muscle activation and limits the control of quadriceps. In midstance the lack of knee extension remains the main gait deviation since it decreases muscle activation in the calf, lower-limb extensors, and hip abductors [16]. In the late stance phase, lack of knee flexion and ankle plantarflexion complicates the push-off and preparation for swing, and is caused by calf muscle weakness [16]. In early and mid-swing phases, the limitation of knee flexion increases stiffness and decreases the activation of hamstrings [16]. Lastly in the late-swing phase, the limited knee extension and ankle dorsiflexion makes the heel contact and weight-acceptance difficult [16]. Stroke often leads to decreased walking speed, short and uneven step and stride length, increased stride width, and increased double support phase [16].

Stroke occurs when a cerebral blood vessel gets ruptured or blocked, which leads to a lack of oxygen from blood damaging the central nervous system cells and tissues [4]. Motor deficits are among the most frequent post-stroke impairments, with approximately 72% of individuals experiencing lower-limb motor dysfunction [7] pp. 71-72. In summary, a stroke frequently causes muscle weakness, which leads to limited knee flexion and dorsiflexion in the affected side. For this reason, the gait appears stiff, slow, and challenging for the patient, which can be eased through walking assisting devices. This is shown by the velocity decrease to approximately a range of 0.18 to 1.03 m/s for patients with post-stroke gait impairments, whereas healthy individuals' average is 1.4 m/s [4]. For stroke survivors, the rehabilitation of gait disorders is a major goal because of the importance walking has in daily activities [15].

The type of gait disorder depends on the location and severity of the brain injury due to the stroke. Hemiplegic gait is a common gait disorder, which is caused by weakness on one side of the body. Approximately 50-85% of hemiplegic stroke patients recover their ability to walk, however their gait patterns are generally seen to be different from healthy individuals [4]. In hemiplegic gait the differences in the stance phase are derived from reduced hip extension in late stance, changes in lateral pelvic displacement and knee flexion, and decreased ankle plantarflexion at toe-off [4]. In addition, hemiplegic gait is characterised by reduced hip and knee flexion during swing phase, decreased knee extension before initial contact, as well as decreased ankle dorsiflexion during the swing phase [4].

Stroke often causes drop-foot, which is an inability to lift the forefoot due to weakness of ankle dorsiflexor muscles, which results in remaining plantarflexion and affects heel strike and toe clearance [17] pp. 399-403. It can be a symptom of a neurological or neuromuscular condition, including hemiplegic gait or steppage gait. It is one of the reasons for an inoperative gait cycle, which is one of the major deficits following stroke [4]. Drop-foot can also occur in individuals with multiple sclerosis or cerebral palsy [18], [12]. In addition to hemiplegic gait, stroke can also cause other gait disorders, for example spastic, ataxic or parkinsonian gait, depending on whether the stroke affects the motor cortex, cerebellum or basal ganglia [19], [20], [21].

2.3.2 Spinal cord injury

Spinal cord injury (SCI) often causes a patient to lose motor skills and the ability to self-balance when walking [22]. The injury usually causes paralysis or weakness of the lower limbs, leading to a loss of voluntary motor control and reduced active joint movements [22]. SCI affects the key muscle groups needed for gait, such as hip flexors, knee extensors and ankle dorsiflexors, depending on the place and scope of the spinal cord lesion [7]. SCI reduces the DOFs in gait, resulting in fewer movement options and reduced adaptability.

Individuals with SCI often experience spasticity, meaning increased muscle tone and exaggerated reflex responses [23]. Spasticity hinders smooth and controlled joint movement, especially during swing and stance phases. Also, impairments in proprioception, awareness of the position and movement of limbs, increases difficulty in adjusting foot placement, further affecting gait. The combination of all these, motor weakness, spasticity, and sensory deficits compromises postural stability, increasing risk of falls during ambulation.

SCI is a lesion that occurs in any allocation of the spinal cord [17]. SCI can outcome as complete or incomplete impairment of motor, sensory, and autonomic functions below the injury level [17]. Complete SCI means that no signals can go through the injury, resulting in full loss of motor and sensory function below the injury level, whereas an incomplete SCI allows some neurological function to remain. SCI can be paraplegic or tetraplegic, with paraplegia affecting the lower limbs, and tetraplegia affecting both the upper and lower limbs. According to literature [17], [3], 250 000 to 500 000 people get spinal cord injury annually [17].

Among patients with SCI, rehabilitation aims to increase levels of independence leading to a better quality of life [17]. The goals also depend on the level of the paralysis. From the patient's perspective, patients suffering from paraplegia pursue regaining the bladder and bowel function, while patients with tetraplegia prioritise recovering the upper limb function [17]. Meanwhile, for patients with incomplete SCI, recovery of walking is rated as the main goal [17].

Roughly 45% of traumatic SCI patients experience incomplete tetraplegia, while 21% experience an incomplete paraplegia, and the rest complete SCI [7] pp. 72-73. Patients with incomplete SCI have a better chance to regain walking ability than those with complete SCI [23], although in European cohort, 8% of all lesion types regained walking ability with one or two braces six months after injury [7]. The requirement for assistance depends on the user's impairment meaning that there is diverse need for different amounts of assistance for gait training and walking support [7].

2.3.3 Parkinson's disease

Parkinsonian gait is a serious symptom of Parkinson's disease (PD), which is believed to have a greater adverse effect on the quality of life than other Parkinson's symptoms [24]. The most notable motor symptoms of PD are rigidity, bradykinesia, impaired postural stability, and rest tremor [13]. The motor symptoms in PD result from a lack of control over movements and difficulty initiating muscle movements [24]. Already at the early stage of the disease, walking often appears considerably slow [13]. The Parkinsonian gait develops with the progression of the disease, which consists of slow gait with a short step length, a narrow base and a leaning posture, reduced arm swing, feet are lifted lower than normally, increased risk of falling, and freezing of gait [13], [24]. Due to these reasons the step-to-step variability of the gait cycle increases [13]. Parkinsonian gait changes can be continuous or episodic [24]. Most of the mentioned features are continuous as they happen all the time while walking. Freezing of gait is an example of episodic changes [24]. Difficulties with gait initiation and freezing often occur suddenly and randomly while turning, changing directions, walking through a crowd, or when approaching narrow passages or obstacles [13] [24]. In freezing the patient loses the ability to pick up their feet, which freezes them in place [24].

PD patients frequently develop a tendency to walk with a forward lean, which is connected to increased step frequency, reduced step length, and bent body posture [13]. Such a gait pattern is called festination, and it includes risk of falling forward [13]. In situations with increased risk of falling, healthy individuals prioritize balance control over other tasks, however with PD patients, this ability is impaired or lost [13]. Additionally, for PD patients, performing other tasks simultaneously worsens the gait, e.g. walking while talking [13].

PD is caused by the dying of nerve cells in basal ganglia which leads to reduced production of a neurotransmitter called dopamine [24]. The basal ganglia forms connections between neurons by using dopamine [24]. Its responsibility is to make the body movements smooth [24]. Less dopamine leads to fewer connections in this area of the brain, so the basal ganglia cannot perform as smoothly as it should, conducting to Parkinsonian gait and the other movement symptoms of PD [24].

2.3.4 Aging

Gait disorders become more common with age. In the elderly, the speed of an individual's walking is a good indicator of general health and survival [13]. Ataxia causes complications with coordinating muscle movements, and is a risk factor for falls [13], [17]. In sensory ataxic gait the stance and gait are insecure and broad based, and the length of the step is shortened [13]. The gait is irregular since the feet are occasionally lifted high leading to the gait possibly having a stomping quality [13]. Also, the cerebellar ataxic gait has similar qualities to the sensory ataxic gait. Ataxic gait and its connected fall risk become worse if the environment is challenging e.g. unknown territory or obstacles [13].

With the elderly, frontal gait disorders are common [13]. Frontal gait disorder appears with patients who have frontal lesions [13]. They often tend to have forgotten the proper act of walking and have challenges with adapting the posture to changing positions and achieving a stable position [13]. Among the elderly, gait disorders usually have several causes [13]. Also in frontal gait disorder, the gait is broad based, step length is short, and anxiousness is noticeable, along with stooped, upright or hyperextended posture [13]. Additionally, cautious gait belongs to common neurological disorders. It refers to an enormous number of changes in walking and fear of falling due to age [13]. Cautious gait often occurs after the first fall and changes the gait to slow, wide based, slightly stooped postured, and reduced arm swing bilaterally [13]. Patients with cautious gait fear falling possibly resulting in intense cases to a complete loss of walking ability [13].

In [25], for two thirds of those strained by any gait disorder the causal actor was neurological, while for approximately half of it was non-neurological, meaning a significant overlap between the two causal actors in patients. The most common non-neurological gait disorders are osteoarthritis and skeletal deformities of the lower limbs [25]. Osteoarthritis-related gait changes result in antalgic gait, but changes in gait due to skeletal deformities depend on the affected area. The most found neurological causes were Sensory ataxia (18%) and Parkinsonian gait (16%) disorders, followed by frontal (8%), cerebellar ataxic gait disorders, cautious gait and hypotonic paretic, spastic, vestibular and dyskinetic gait disorders [25]. In the study, roughly a third of the patients had multiple neurological causes for their gait disorder, increasing the difficulty of a precise classification.

The gait disorder frequency reaches over 60% in people over the age of 80 years, while being only 10% in people aged 60 to 69 [13]. According to [14], more than 80% of people aged over 85 have a gait abnormality. In the coming decades, there will be an increased frequency of gait disorders originating from the expected demographic changes [13].

3 Technical designs of exoskeletons

Exoskeletons are assistive devices that are used to strengthen the user's weak muscles to recover the locomotion capacity [8]. They are mechanical structures that imitate lower limbs and are designed to help the patients [22]. As was mentioned in the previous chapter, joint motion, muscle control, and balance are essential factors for a properly functional gait. The challenges presented with the impairment of said factors define the functional requirements which exoskeletons need to meet using their control systems. Accordingly, exoskeletons need to support transition between stance and swing phases, weight shifting and coordination correspondingly to the gait cycle. Beyond their assistive function, exoskeletons are also generally used for rehabilitation purposes. During the previous decade, rehabilitation lower-limb exoskeletons have become essential because of the fast increase in amount of neurological disease patients [22]. The main uses of exoskeletons include enabling movement and providing stability [22].

Exoskeletons can be divided into assistive and rehabilitative exoskeletons. The design goal of assistive exoskeletons is to aid impaired people with performing daily activities. This is achieved through improving mobility, reducing fatigue, and enhancing independence in tasks including walking or standing. Rehabilitation exoskeletons, on the other hand, are used in clinical settings to improve motor recovery through repetitive, guided movements. These devices aim to stimulate neuroplasticity, improve gait patterns, and restore function over time [22]. Assistive exoskeletons are typically designed for long-term support, and they follow the user's volitional movements, whereas rehabilitation exoskeletons are focused on motion patterns and functional recovery.

Rehabilitation lower-limb exoskeleton design structure can be divided into overground and immobile devices [22]. Roughly divided, overground exoskeletons are assistive by design while immobile exoskeletons are more rehabilitative. Figure 3-1 displays three different exoskeletons: the first two are overground full lower-limb exoskeletons, and the third one is immobile treadmill exoskeleton. Overground devices are utilised with patients able to walk and stand when assisted with stability and mobility [22]. There are two operation modes in overground exoskeletons: body weight-supported and unsupported body weight [22]. Immobile exoskeletons are used for patients with difficulties with standing and walking on their own due to various disorders [22]. For immobile models the walking area is restricted in contrast to the overground models [22]. Immobile exoskeletons can also be divided into two types: fully assisted mode, and assist-as-needed mode [22]. In the first, active participation of the patients is not needed in the movement [22], whereas, in the second, the effort of the patient towards movement is complemented by the exoskeleton's appropriate motion [22].

Exoskeletons aim to grant freedom of movement to people with physical impairments by providing the joints with additional locomotion strength [8]. Standing and walking may improve multiple aspects of health, regardless of it being done independently or assisted [9]. According to literature, combination of robotic and conventional rehabilitation training positively influences independent walking, walking capacity, and walking speed [26]. Additionally, there is an understanding that gait performance can be improved by guiding movements although the user is passive, especially with patients suffering from severe impairments [26]. According to [26], no solid proof of robotic rehabilitation being superior to conventional therapy has been found. Although, [27] showed that post-stroke subjects performed better in their motor recovery, walking endurance, and ability to balance and walk after six weeks of partial body weight support rehabilitation four times a week.

Soft exoskeletons, also referred to as exosuits, are a new generation of wearable assistive devices that are different in both structure and function. Soft exoskeletons are developed using flexible, compliant materials that adapt to the user's body, offering improved comfort, lighter weight, and reduced intrusiveness compared to rigid exoskeleton designs [22], [28]. These devices are designed to support movement without restricting natural mobility. Soft exoskeletons help reduce misalignment issues, which increases the comfort of use [22]. The technology aims to achieve more natural walking dynamics through diminishing gait disruptions [28]. Soft exoskeletons, exosuits, differ from rigid exoskeletons by requiring the ability of independent walking [28]. This leads to the user base consisting of elderly or poststroke individuals, who are not in need of significant assistance [28]. With the promising innovations in soft exoskeleton technology, incorporating its use with traditional exoskeleton designs is likely to improve weight, comfort, and strength [28]. However, as this thesis is focused on rigid exoskeleton designs, soft exoskeletons will be left with a suggestion for further research on its applications.

Exoskeleton is a powered device that uses a system of actuators and sensors to execute walking movement [9]. This chapter introduces the technical side of the exoskeletons, which gives a good understanding of exoskeleton limitations and fundaments the reasons for current exoskeleton design problems. There is also an introduction to different types of exoskeletons to facilitate understanding of diverse exoskeleton aims and models.

Figure 3-1. On the left side HAL lower-limb (HAL-ML05) exoskeleton [29]*, in the middle ReWalk 7 exoskeleton* [30]*, and on the right side a treadmill Lokomat exoskeleton* [31].

3.1 Mechatronic of Exoskeletons

Exoskeletons' mechatronic structure consists of actuators, power supplies, sensors, and controllers [22]. Exoskeleton joints are driven by motion converted from energy by actuators, which are small mechanical devices. The exoskeletons' mechanical properties must correspond with human biomechanics. Thus, features such as compliance, linearity, torque, shock load effect, backlash, compactness, reliability, controllability, and noise need to be considered in the actuator and power transmission mechanisms [22]. Actuator selection and power transmission design are important in rehabilitation exoskeleton development, as they have a strong influence on motion performance and human-machine interaction [22]. Also, the location of actuators needs to be considered since it directly affects the mechanical design and the control architecture of the exoskeleton [32]. The actuator system should adapt to the human biomechanics as it composes the power transmission mechanisms [22]. Therefore, the actuator and power transmission mechanisms play a key role in ensuring the effectiveness, safety, comfort and smoothness of exoskeleton motion [22]. The actuators are controlled with various control strategies, which are integrated to provide stability to the user even in emergency situations [8]. The sensor measurements are the base of exoskeleton operation as they provide the essential information from the current gait phase of the cycle and the user's position to the control unit. Understanding the mechanical design is critical for exoskeleton development.

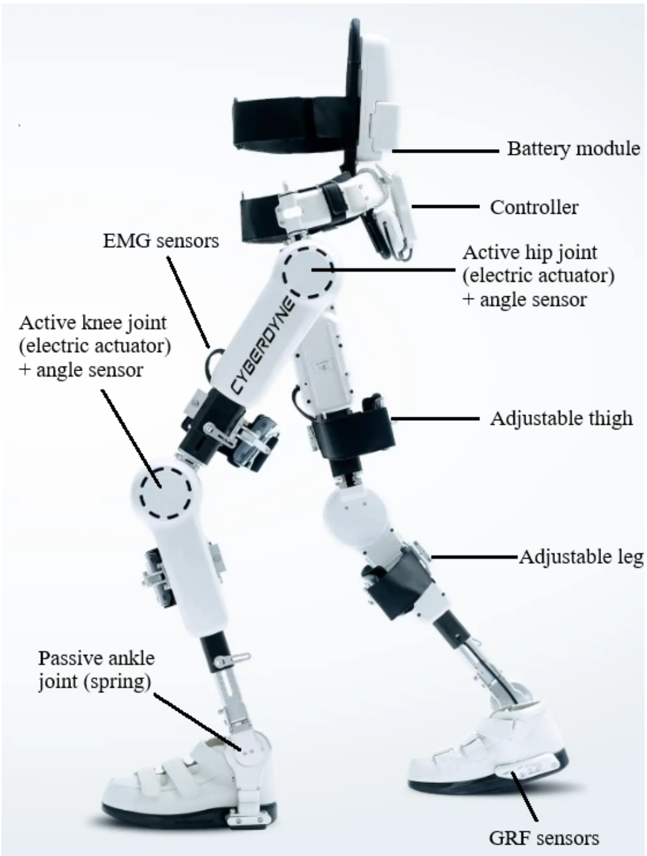

Figure 3-2. Actuators of HAL exoskeleton (base picture from [29]*).*

3.1.1 Structural Components

The mechanical structure of exoskeleton actuation systems should be designed to mirror the musculoskeletal system to ensure comfortable and effective human-machine interactions [22], [32]. It is important that the exoskeletons' mechanical design matches the human lower-limbs for the exoskeletons to synchronise limb movement and mimic a standardised gait [22], [32]. Exoskeleton lower-limb joints can be divided into multiple joints and single joints [22]. In multiple joints, more than one joint (hips, knees, and ankles) is actuated, whereas a single joint is actuated with only one joint [22]. The selection of the number of DOFs for the exoskeleton can assure the kinematic alignment with the user's limbs, enabling maximised safety, minimised collision and comfortable walking [32].

The structural design of the exoskeleton depends on whether the device is intended to assist mobility in daily life, provide rehabilitative aid in clinical settings, or support with standing or walking stability. In addition, the design is affected by the joints the exoskeleton is intended to support. Structural decisions are in connection with the nature of the user's impairment and the biomechanical functions that need to be restored. For example, in the case of drop-foot, where dorsiflexion is affected during swing phase, a targeted ankle exoskeleton may be suitable. In contrast, patients with more complex motor deficits, such as a stroke or SCI, may require a multiple joint exoskeleton that supports hip, knee and ankle. The biomechanics is in a central role when selecting an appropriate exoskeleton that matches both the functional and structural requirements of the user.

The exoskeleton's mechanical structure primarily determines user alignment and comfort, as it defines how well the device adapts to the user's body and joint axes. Misalignment results from the unsuitability between rigid structures, e.g. frame, of exoskeletons with the human anatomy, possibly causing undesired translation forces with even a slight misalignment [22]. Due to this, a mechanism and configuration which properly aligns with the joint structure of humans should be developed. Misalignment can result from three main sources: shifting of the joint axis during movement, incorrect alignment while the device is worn, and poor anatomical replication of human joint structures in the exoskeleton design [22]. These misalignments can cause unwanted interaction forces, which may restrict the user's movement and lead to discomfort, pain, or even injury [22].

Misalignment can be reduced with proper fitting and different kinds of adjustable joint mechanisms. Examples include Schmidt coupling for knee alignment, parallel self-alignment designs in the ankle, three DOFs achieved with sliders and hinges, and slot-based length adjustment structures [22]. Length adjustability is particularly important for accommodating the wide variation in user heights and sizes, as it supports better anatomical alignment between the user and the exoskeleton. For instance, in a knee-ankle-foot exoskeleton prototype [7] p. 169, a connecting linkage adjusts the length of shank from 40 to 47 cm, and the HAL5 exoskeleton is available in three sizes (S, M, L) to accommodate different users [1]. According to literature, exoskeletons are generally suitable for 160-190 cm in height, and the maximum weight limit is 80-100 kilograms [1] , [8]. However, there are some exoskeletons that suit individuals between the height of 150-195cm. Designing an exoskeleton that would fit all potential users remains challenging. Although length adjusting mechanisms can be used, it is almost impossible to design an exoskeleton that fits all body types. In practice, proper fitting might still require a trained technician's aid, especially with the initial setup.

The selection of structural materials in exoskeleton design plays an important role in determining weight, durability, comfort, and overall usability. Lighter materials help reduce fatigue and improve alignment of the device. Also, in daily living environments the durability of the material is important as it needs to be robust against suboptimal circumstances, like dirt, humidity, and misuse [22]. According to literature [32], [7], different construction materials of exoskeletons are steel and aluminium mixture, aluminium, titanium and carbon-fibre. Lightweight composite materials, such as carbon-fibre or plastic, meet the ergonomic requirements [7]. Steel and aluminium alloy materials ensure robustness and lightness of the exoskeleton [32]. For example, HAL frame composes from the steel and aluminium alloy materials [32]. Aluminium and titanium are lightweight materials [32], ensuring portability of the exoskeletons.

3.1.2 Actuation and Power

In exoskeleton design, the selection of the actuation type is important, as it influences performance [22]. When selecting actuators, two main factors need to be considered: the structural types and actuator properties [22]. The structural types determine the importance of actuator factors like weight, energy efficiency, precision, and compliance, based on the mode of assistance and mobility setup [22]. Actuator properties include features, such as speed, torque, weight, and power consumption [22]. The structure type defines strongly the actuator selection [22].

Critical compliance features are not required in the actuators of immobile exoskeleton in fully assisted mode, instead, precise position control is [22]. However, both precise actuator positioning and compliance feature requirements are important in the assist-as-needed mode [22]. The safety and softness of actuator motion are important to take into account in both modes, but the weight, compactness, and energy consumption are not critical issues for either in actuator consideration [22]. Differently, in overground exoskeletons the actuator compliance, precise positioning and softness characteristics should be considered, as the user has more control over their movement and provided assistance [22]. The energy storage is also limited, so the energy consumption and compactness of actuator systems are important to consider [22]. As a result of these, immobile exoskeletons are bulkier and heavier than overground exoskeletons, which can be seen in Figure 3-1, as it needs to be able to fully support the body. They also have strong motors, since they can have an external power supply and often require more torque. The torque requirements guide the exoskeleton selection as the person's ability to generate torque depends on the level of impairment.

Actuators can be divided into two categories including active and passive actuators [22]. Active actuators can again be divided into four actuator systems, including electric, pneumatic, hydraulic, and hybrid actuators, which provide mechanical power for exoskeletons [22], [32]. However, this thesis focuses primarily on electric actuators, as pneumatic and hydraulic systems are considerably less common in rehabilitative and assistive exoskeletons due to their weight, complexity, and maintenance requirements.

Every active actuator has two types of motion: linear and rotation [22]. According to literature, electric actuators are the most common type of actuators, as they are lightweight, easy to control, silent, compact, reliable, and offer high accuracy and specific power [32], [6]. In addition, they are low in cost, consist of clean actuation systems, have high precision, have a good power-to-weight ratio, are easy to install and control, and they adapt easily to multiple DOFs in all planes, which are all high in demand [22].

Electric motors are a promising consideration for the portability of exoskeletons while differentiating as the better option from fluidic-based actuators [22]. Unlike hydraulic and pneumatic actuators, electric actuators have no issues with possible fluid leakage [22]. However, electric actuators require a power transmission mechanism to deliver torque from the motor to the joint. Common solutions include gear system, which can reduce rotational speed and amplify torque at low speeds, as well as tendon-driven mechanisms, Bowden cables, ball screws, and belt systems [22]. Actuator positioning influences the efficiency and mechanical design of power transmission [22]. Motors can be placed directly at the joints simplifying routes for transmission but increasing distal weight and possibly affecting comfort of use and control. Another placement option is remotely, which uses tendons and cables to transmit torque. This reduces weight on the limb but introduces possible losses or delays in force transmission. In addition, actuator placement and characteristics of the transmission mechanisms can also affect ergonomics due to their influence on weight distribution and mechanical resistance [22]. Major challenges of actuators and the transmission mechanism are seen in misalignment between the exoskeleton and the patient's joints [22].

Batteries are often used as a power source in exoskeleton devices. They provide a convenient power source for such devices, enabling portable and quiet operation that results in user comfort [22]. Batteries and the main computer are in a backpack in many exoskeletons, which makes it heavy and bulky [28]. However, they are located in the centre of gravity, which is the ideal location to avoid unevenness of movement and weight [28]. Improving energy efficiency helps increase autonomy and reduce battery size, which is a common challenge in electric motor technologies [28]. HAL5 exoskeleton weighs approximately 14 kg and can operate one hour of continuous use [29]. In comparison, ReWalk 6.0 exoskeleton weighs approximately 23 kg, and its operation time is over four hours [30]. According to [8], an average weight of reviewed full lower-limb exoskeletons is around 18.5 kilograms. There is a need to develop some advanced arrangements that can accumulate and supply the energy for various phases of the gait pattern [8].

Series elastic actuators (SEAs) combine a stiff actuator, such as an electric motor, with an elastic element like a spring to achieve compliant motion [7] p. 143-161. This configuration mimics the muscle-tendon structure as it stores and returns elastic energy corresponding to muscles. SEA is a good choice for lightweight and wearable exoskeletons, as it reduces power demands, and enables compact and safe design [7], which will eventually lead to smaller batteries.

An alternative solution is passive actuators, which do not require electricity, but operate by storing potential energy [22]. They are usually lighter and cheaper than active actuators [22] and used in joints that do not require a great amount of assistance. When considering the usage of passive actuator implementations, certain factors should be taken into account [22]. For example, a joint angle analysis should be conducted, as certain angles play a significant role [22]. Passive actuators are often combined with active actuators within the same exoskeleton system, which is called a hybrid actuation model. Hybrid actuators combine two or more actuation principles to enhance performance. Assistive and rehabilitative exoskeletons typically include an electric-motor-spring configuration, which aims to reduce mechanical impedance and improve energy efficiency [22]. Spring-based hybrid actuators achieve light weight and responsiveness by storing and releasing energy at certain phases of the gait cycle. Such designs aim to find a balance between torque output and compliance without compromising the size or complexity of the design. For example, HAL-exoskeleton has two actuated DOFs for each leg at the hip and knee joints and one passive DOF with spring actuation at ankle joint (illustrated in Figure 3-2), to reduce power consumption and weight [22], [32].

Exoskeleton structures have a different amount of DOFs, depending on the goal of actuation and rehabilitation. The number of DOFs is mainly dependant on the joints located in the hip and the ankle [28]. Frequently, knees and hips extension/flexion movements are implemented with active DOFs as they are both essential to develop a normal gait [28], whereas hip ab/adductor and ankle dorsi/plantarflexion are often implemented with passive actuators [28]. Actuated ankle improves the balance capability, however, the control complexity increases in addition to weight and size [28]. Another reason for passive actuators in the ankle joint is their location at the end of the leg [28]. Active actuators lead to increased inertia and torque needed by the knee and hip actuators, as well as the user's increased sensation of weight [28].

3.1.3 Sensors and Control

The interaction between the user and the exoskeleton is realised by sensors and control systems. The control system uses the sensor data in order to interpret user intent and deliver movement assistance accordingly. The control system aims to assist users in maintaining normal gait patterns, improve ambulation, and aid rehabilitation [32]. The effectiveness of an exoskeleton depends on how well its sensing and control systems match the user's specific impairments and rehabilitation goals [33].

To understand what role sensors play in gait analysis, notable measurements need to be identified, and the sensors applied to collect them. A key target of sensing is understanding the user's gait rhythm which includes step time, cadence, swing and stance durations etc. as introduced in chapter 2. Gait rhythm can be quantified by analysing joint angle profiles and patterns of joint movements over time to identify gait cycles and their phases. The functional roles of sensors depend on their application. Stability systems are used to detect disruptions in balance, whereas rehabilitation exoskeletons need sensors for precise gait phase detection and progression. These measurements can be achieved through sensors such as inertial measurement unit (IMU), force-sensing resistor (FSR), and ground reaction force (GRF).

IMUs are among the most common sensors used in wearable exoskeletons. They use a combination of accelerometers, gyroscopes and sometimes magnetometers and temperature sensor for the measuring systems [34]. IMU sensors gather the kinematics of the gait [7] p.

172, measuring the body's movements, acceleration, and angular rate [34]. They are commonly used for gait phase detection, pose estimation, and intent recognition [7], [26].

FSRs measure pressure or force applied to specific areas of the foot. They are frequently used in supervisory-level controllers to detect gait events by identifying foot control and lift-off phases [32]. GRF sensors measure how the body interacts with the ground during walking or standing [26]. This allows accurate detection of gait phases of the foot contact and lift [26]. Pressure sensors on the insoles can also help estimate the user's Centre of Mass, Centre of Pressure, or Centre of Gravity [28]. FSRs are useful for contact detection and force estimation, however the more expensive GRF systems are more accurate and comprehensive in detecting and measuring force. FSR and GRF sensors are usually placed at e.g. heel, toe, and metatarsus regions (as illustrated in *Figure 3-2)* [26].

Potentiometers, encoders, and hall angle sensors are used to obtain joint angle measurements which are fundamental in exoskeleton control [7], [6]. Joint angle feedback helps detect gait events and supports position control [6], [7], especially in individuals with hemiplegic gait [26]. In some devices, gyroscope placed on the backpack detects falling [7] p.195.

Electromyography (EMG) sensors are used in advanced exoskeleton models to measure muscle activity and can be used to assess user intent [28]. The combination of EMG together with other sensors, such as joint angle and plantar force, provide more accurate control response [28]. Electroencephalography (EEG) sensors, while mainly used in research and not commercially, are being researched to capture movement intent straight from brain activity [28]. While EMG and EEG provide valuable data, they can complicate device donning and everyday usability [28]. However, their integration allows faster and more natural user-device interaction benefiting the walking ability [28].

As introduced, the selection of an exoskeleton requires a comprehensive understanding of sensing systems but also the control algorithms each device utilizes [33]. Although sensors collect the raw data from the user's movements and physical condition, the control strategy actually interprets the data and decides the response of the system. In other words, the effectiveness of assistance depends not only on what is measured, but also on how that information is processed and acted upon. Control strategies transform sensor input into real-time motion assistance.

Exoskeletons apply various control strategies to improve user mobility and rehabilitation. The control of a lower-limb exoskeleton includes synchronizing robotic assistance with the user's movement, intension, and physical condition, compiling into a complex task. Assistive and rehabilitation exoskeletons both have the control system defining the motion and interaction of the body and the device safely, effectively, and adaptively [26]. The controllers' purpose is to support motor function delivery by precise assistance and resistance during movement [26]. This is done by generating appropriate joint trajectories, applying supportive torque, and modulating stiffness based on the user's state and environment [26]. The main goal control in exoskeletons attempts to achieve is providing timely and personalised assistance which helps facilitate walking without overcompensating the effort of the user or disrupting their natural gait patterns. To achieve this, a combination of biomechanical modelling, sensor integration, and real-time decision-making are required [26]. Additionally, the exoskeleton is able to store energy for various applications through the optimization of the control strategies and the gait pattern [8].

Main variables to be regulated include joint position, joint velocity, torque, and stiffness. The sensors, described earlier, track these key biomechanical parameters in real time. As described in chapter 2, exoskeleton users have highly varying movement, both between individuals and between steps. This makes precise predicting and matching of motion patterns difficult for the controller. Due to the high variability, it is important to have real time data input from the sensors for the control system, since the exoskeleton control operates continuously. Any significant delay between user action and device response can compromise safety, disrupt balance and lead to risk of falling. With the challenges sourcing from real-time regulation and movement variability, exoskeletons rely on a layered control system to ensure accurate and responsive assistance.

Exoskeleton control systems are commonly hierarchically divided into high-, mid- and low-level control [26], each responsible for a specific aspect of decision making and actuation. High-level control identifies the user’s intensions and selects corresponding exoskeleton response behaviour [26]. Commonly used high-level strategies include Finite State Machines, which detect discrete walking phases and select appropriate control responses based on the phases [32]. Human-robot synchronisation is a sub-level of high-level control, which estimates the condition of the user by using data-driven methods informed by multiple sensor modalities [26]. High-level control can also be described as assistive or challenge-based. This description depends on whether the aim is to support the movement of the user or intentionally require the active effort of the user to facilitate motor learning. Mid-level control translates the user intentions into movement commands [26]. It adjusts the exoskeleton’s target trajectory or force according to the user’s estimated state and the intended control strategy [26]. This level often applies adaptive control strategies to allow real time assistance modification for the system based on user performance or physical conditions. Low-level control aims to accomplish the desirable state determined by the mid-level controller [26]. It executes the precise joint movements, and correspondingly they are straightly connected to the hardware [26]. Position and torque control based on PID or PD are the most used low-level controllers [32].

Based on the hierarchical control structure, various control strategies are implemented at each level to maintain a proper human exoskeleton interaction between the user and the device [8]. Therefore, establishing a smooth control strategy is important to promote the accuracy, effectiveness, and comfort of the device depending on the user’s intentions [8]. In the literature [8], nine different control strategies are introduced: Reference trajectory-based gait control, model-based stability control, adaptive oscillator-based control, predefined motion-based control, sensitivity magnification control, decision based fuzzy control, hybrid assisted control, muscle stiffness control, and proportional EMG-based control [8]. These control strategies are presented shortly in Table 3-1.

Table 3-1, Control strategies introduced according to the control level. Function descriptions for each control strategy are compressed from [8], [26].

Control strategy	Control Level	Function Description
Reference trajectory-based gait control	Mid-level	Uses pre-recorded reference joint trajectories for assistance in individuals with impaired mobility.
Model based stability control	Mid-level	Uses zero moment point and centre of gravity methods to stabilize movement.
Adaptive oscillator-based control	Mid-level	Synchronizes the movement of the exoskeleton with that of the user's periodic locomotion-related signals.
Predefined motion-based control	Mid-level	Executes movement using pneumatic or spring-based actuators based on estimated gait timing.
Sensitivity magnification control	Mid-level	Enhances small user movements to generate full exoskeleton action, used in mutual force-interaction systems.
Decision-based fuzzy control	High-level	Makes high-level control decisions based on interpreting inaccurate input using fuzzy logic.
Hybrid assisted control	High- and Mid-level	Combines multiple control strategies simultaneously during different gait phases.
Muscle stiffness control	Low-level	Adjusts stiffness of joint via models based on musculoskeletal dynamics. Mainly implemented on the knee joint.
Proportional EMG-based control	Low-level	Converts myoelectric impulses into motor actuation.

Reference trajectory-based gait control, Model-based stability control, Adaptive oscillator-based control, and Predefined motion-based control are common control strategies which are used in both single- and multi-joint exoskeleton systems [8]. Sensitivity magnification control, Decision based fuzzy control, and some Hybrid assisted control strategies can only be found in multi-joint systems, while Muscle stiffness and Proportional EMG-based control are only seen in single-joint exoskeleton systems [8]. The most common strategies used for the mid-level control are position trajectory tracking and reference trajectory-based control in the exoskeletons reviewed in the study in [26]. Trajectory-tracking is well utilised in regular and controlled environments but can be seen having difficulties when facing varying gait or irregular user behaviour [26]. In contrast, compliant control (muscle stiffness control and impedance control) improves the comfort of the user by regulating support based on motion errors, although this may reduce precision [26]. According to the review study, the combination between trajectory-tracking and compliant control showed the highest clinical effectiveness, however it also requires the longest time of training. Adaptive control strategies portray more advanced systems that are able to modify parameters in real-time based on the user's condition or performance [26]. This comes with the downside of requiring extensive sensing and computational resources for effective functioning.

In the case of HAL exoskeleton, the state machine resolves in which walking pattern phase the user is, using the information from the sensors [28]. Different information and strategies from various sensors are used, depending on the mode of the exoskeleton [28]. HAL has three different modes, which are: Cybernic Voluntary Control, Cybernic Autonomous Control and

Cybernic Impedance Control [28]. These modes can be linked to the general control strategies which were presented in Table 3-1, although HAL's application is distinctive in combining multiple approaches. For example, Cybernic Voluntary Control utilises EMG signals when detecting the user's voluntary movement intent. This does involve interpreting myoelectric input, however, it does not follow a plain proportional EMG-based control strategy. Instead, initiating and modulating assistance through sensitivity and balance parameters is done with EMG, simulating sensitivity magnification control where small voluntary efforts are amplified by the system [28]. Cybernic Autonomous Control uses postural and plantar pressure measurements to define the desired motion [28]. It corresponds with predefined motion-based control and occasionally reference trajectory control, based on how the motion is produced. Lastly, the Cybernic Impedance Control supplies torque to compensate for the actuator's resistance [28]. This mode reduces the resistance of the actuator enabling free movement without generating additional torque to assist with the user's motion. The various modes portray the adaptive capabilities of exoskeletons to meet changing user needs and rehabilitation goals. Although the basic idea where voluntary and autonomous control systems are combined is a common practice in advanced exoskeleton systems, the modes of HAL work as a unique example of implementation [35].

The goal of many control strategies is to automate exoskeleton function through sensor-driven algorithms, however, current sensing technologies may sometimes not be sufficient enough to capture the user's intent accurately and in real time [8]. Therefore, some devices do not use sensor data for control, but supply the user with a joystick for control of the exoskeleton [36]. In addition to a joystick, a hands-free exoskeleton can be controlled with buttons or a band, similar to a watch [28], such as REX or ReWalk exoskeletons [36], [28]. Also, exoskeletons with crouches or canes can be equipped with buttons [7] p. 195.

Although current control strategies aim to meet the requirements of an ideal controller, they often face limitations and challenges when applied to real users and environments. These limitations can be categorised into four main areas: user-related variability, sensor limitations, real-time computational constraints, and environmental variability. Humans are individuals so there are differences between users, for example different gait disorders lead to divergence of muscle strength and range of motion. Resulting from this, the same control settings may not be implemented for all individuals, leading to a need for personalization or adaptive control. Sensor-related limitations can lead to total failure of the exoskeleton system if even a small variation or disturbance occurs in the input signal [32]. Sensors may not deliver the data quickly enough to keep up with real-time requirements, resulting in delayed control responses. Also, efficient computation is required by real-time processing of control algorithms to maintain responsiveness [26]. Sensor accuracy, such as IMU or EMG, can be unstable due to sensor noise or signal variability [26]. IMU sensor output accuracy decreases the longer it operates, as the sensors drift over time. EMG signals lack robustness due to variability between patients and dependence on sensor positioning [26], resulting to skin impedance changes. Surface EMG sensors are common due to their non-invasive nature, but intramuscular or intravascular EMG has also been proposed to capture more precise signals and to solve this challenge [8]. Furthermore, all sensors may occasionally cause error signals, causing the possibility of wrong operation of the controller and actuators. In addition, environmental factors influence the gait of individuals with exoskeletons. Gait can change due to uneven surface, inclines, obstacles, or turns [13]. The exoskeleton’s control system must be able to react quickly to swiftly changing situations, as an obstacle on the ground can cause a stumbling increasing the risk of falling [13]. Delays, system errors, or lack of intuitive feedback may cause mental fatigue compromising user performance and engagement further.

4 User-Centred Design approach

Exoskeleton users are often divided into primary users, secondary users, and tertiary users. When addressing exoskeleton users, it often refers to primary users (PUs). They are the ones who actually are using the exoskeletons and directly profit from the devices [7] pp. 71-76. PUs include people with e.g. SCI, stroke, and elderly people with mild-moderate mobility impairments [7] p. 2. Secondary users (SUs) are those who have direct contact with PUs [7] pp. 71-76. They are e.g. doctors and physiotherapists, caregivers, spouses, family, and friends [7]. Tertiary users (TUs) refer to stakeholders that do not use the device directly but influence its implementation, for example through regulation or funding decisions [7]. They include e.g. insurance companies, regulatory agencies, hospitals, rehabilitation centres, standardisation bodies, and advocate groups [7] p. 2. Tus are often not part of the main consideration in the development of new devices, although they too need to be regarded as stakeholders [7] p. 75.

The biomechanics of gait and functional gait cycle together with mechanical aspect of exoskeletons are important to understand in order to design exoskeletons according to user needs. Today exoskeleton use is mainly for health and rehabilitation purposes while supervised in clinical settings but is to be developed to facilitate daily use as a functional mobility device [9]. Exoskeleton design is guided by the device’s goal. Therefore, the user requirements depend on the type of the exoskeleton. The use of exoskeletons in rehabilitation is growing. First exoskeletons start to get their approvals for home use, so it is likely that the exoskeletons will be increasing in the future both in clinical and home settings. For the future improvements in exoskeleton design, four key topics have been identified: robust control, safety and dependability, ease of wearability or portability, and usability and acceptance [9]. With any of these problems, it is likely that the exoskeleton is abandoned nor is it used to its potential [9]. Therefore, the end user must be central in the design and development process of the technology [9]. For exoskeletons to be accepted and used by users, they should follow user centred design principles.

In user centred design (UCD), the aim is to reach and manage the entire user experience [37]. The process starts with precise understanding of the users, tasks and environments [37], in this case, the user’s disorder and functional need for the exoskeleton. UCD is an iterative process in design, where each design process phase is focused on [37]. There are four phases in each iteration of the UCD approach [37]. The phases are understanding the context of use, identifying and specifying user requirements, design solutions, and evaluation [37], which are illustrated in Figure 4-1. In the last phase, the outcomes of evaluation are valued against the user’s context and requirements, to see how well the design is performing [37]. This iteration is continued until the evaluation results are pleasant [37]. UCD involves users in the design process, hence the products, in this case exoskeletons, are better positioned to satisfy user expectations and requirements [37].

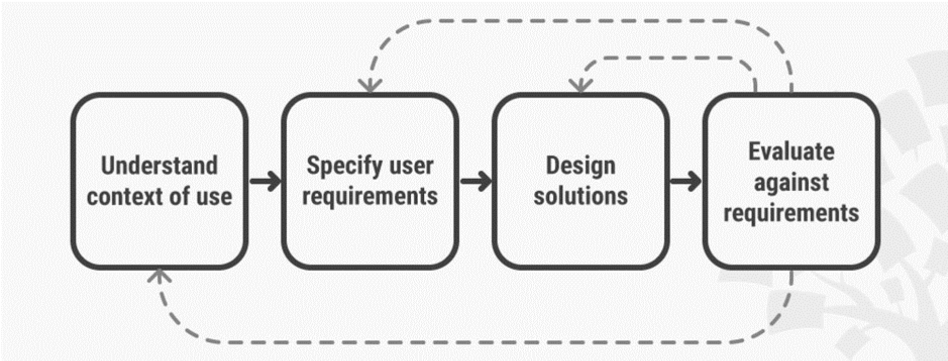

Figure 4-1. User centred design is an iterative process following four phases seen in the figure. The figure is adapted from [37].

In this chapter, user requirements for overground exoskeletons are examined, with particular attention to current limitations derived from mechanical design or misalignment with human biomechanics. Immobile exoskeletons, typically used in clinical settings, are less demanding in user requirements. Their common UCD challenges include fitting and cost, as the exoskeletons are intended for multiple users with varying body sizes. Adjustable sizing is therefore important, as it also reduces the cost by decreasing the need for individualised devices. Although, these fitting and cost procedures apply to both immobile and overground systems, overground exoskeletons have more complex demands for daily usability and personalization.

Overground assistive exoskeletons can enable daily activities for impaired patients together with enhanced possibility for rehabilitation. One of the issues in exoskeleton usage is that they are not generally suitable for home use. Home usage allows more effective rehabilitation and increases the health benefits as the user can use more of the device. Although, overground exoskeletons are assistive devices, they are currently used primarily as rehabilitation tools, e.g. HAL and ReWalk for SCI patients [7] pp.72-73. However, health benefits and social interaction may stand out more for the overground exoskeletons due to the narrow potential users benefiting from rehabilitative effects [9].

The exoskeleton features for assistive exoskeletons have been studied widely. In a study by [9], participants consisting of wheelchair users and healthcare professionals, rated 17 potential design features from 1 (Very Unimportant) to 5 (Very important). Majority of the participants gave a rating of five to four of the design features: comfort, price, minimizes risk of falling, and putting on and taking off the device. Also other features were rated as important by both of the participating groups by approximately 72% of participants: ability to walk on uneven surfaces, range of battery life, amount of energy needed for use, device portability, and ability to carry out daily asks while standing [9]. Also, in [7] pp. 83-84, the PU and SU design requirements and must-have features are assessed. Both user groups quantified easy to don and doff, improve safety and stability for users to prevent falls during gait and transfer movements, lightweight, easy to use, low cost, and compatible with user footwear and clothing as important features [7] pp. 2-3.

SU consideration in exoskeleton development is significant, since it gives insight into the support and remote interference needed from healthcare professionals and caregivers through the life cycle of the product [7] p. 69. In many assistive exoskeletons, the setup, customization, and maintenance of the device are often carried out by SUs [7]. Therefore, their needs can often be seen as important as PUs [7]. For example, ReWalk exoskeleton is approved by the United States Food and Drug Administration when a specially trained assistant is assisting the user [9]. The need of assistant restricts the independence and makes the device less accessible as it is difficult in home use.

According to [9], affordability, comfort, safety, and ease of use are required to improve to achieve users goals, before exoskeletons can be adapted as mainstream mobility devices. Those features arise continuously in literature. In addition, exoskeleton use also influences cognitive and social aspects, which are increasingly recognised as important in recent studies. Therefore, all of these themes are examined more deeply in context of daily exoskeleton usage.

4.1 Cost

Cost is one concern to potential users of exoskeletons. The high cost of the devices results in clinical centres or hospitals with low budget to not have access to them, thus being even more challenging for personal use [28]. In the study [9], healthcare professionals and wheelchair users were asked to propose a price range appropriate for an exoskeleton. The median price, reported by healthcare professionals, was $10,000- $20,000 USD, whereas for wheelchair users it was under $10,000 USD. Purchasing e.g. a ReWalk exoskeleton for personal use costs almost $70,000 USD, which differs significantly from the reported acceptable cost [9].

The price of the exoskeleton depends on the type, size and customization of the device. The price increases in correlation with the level of actuation. Ankle-foot exoskeletons are significantly cheaper than full lower-limb exoskeletons as they require less actuated joints. Bigger exoskeletons also require more torque, meaning more actuators, sensors and bigger batteries, resulting in an increased price. In addition to a high purchase cost, stakeholders also have concerns with maintenance costs [9]. Exoskeletons require regular maintenance, including hardware service, calibration, and component replacement, larger and more complex devices requiring them even more frequently. Full lower-limb devices with multiple joints, actuators, and sensors require more frequent servicing than simpler assistive systems. These maintenance demands increase the total cost of exoskeleton ownership. In addition, some exoskeletons also require a trained assistant while using the exoskeleton (e.g. ReWalk). Continuous need for an assistant may also increase the costs significantly.

Customization of the device also increases the cost of the exoskeleton, as the size and level of actuation is tailored for the user. Fully tailored exoskeletons deliver superior fit and comfort but are more expensive and less scalable. Therefore, adjustability of the device is important, as the system supports multiple users, but they require more complex and adaptable components. Adjustability can reduce the price and is favoured especially by clinical or shared environments where user needs vary.

With assistive exoskeletons, the users can be more independent which reduces assistance needs by the user consequently reducing care costs [7] p. 75. Also, rehabilitation with exoskeletons can speed up the healing process, lowering the overall treatment cost [7]. With assistive devices, users can be active in society, including working. Exoskeletons allow patients to increase their employment, which would benefit the whole economy.

Exoskeletons are very expensive to purchase and maintain, with not many being able to afford them [8]. Due to the high cost, there is a need to develop an affordable exoskeleton, so the maximum amount of elderly and impaired people can benefit from them [8]. Integrating new technologies for actuators, batteries and sensors can reduce the price of exoskeleton systems [8].

4.2 Comfort

The comfort of using an exoskeleton is an important user requirement, as it partially defines whether the user will use or abandon the device. In order for an exoskeleton to be comfortable to use, it needs to be portable, lightweight and compliant. Additionally, easy transportability, having improved dynamic stability, user interface, and terrain navigation should be accomplished for home mobility users to be able to walk confidently with little to no assistance in their daily life [7] p. 368. For achieving portability, weight continues to be significant issue. Exoskeletons are often heavy with limited torque and power, which makes the wearer’s movements difficult to augment [18], [8]. Current exoskeletons are bulky, unnatural and noisy, and additionally could be personalised better [1]. Bulkiness is caused by the weight and size, which consequences from the frame and actuators of exoskeletons. New designs or improvements of actuators are needed with characteristics, such as lightness, compactness, compliance and lesser energy usage [22]. Also, the material should be chosen accordingly to achieve more lightweight and easily portable designs [8]. Material choices should also support convenient breathability, as heat and sweating can reduce comfort, especially in bulky or enclosed designs.

Current exoskeletons often cause discomfort to the wearer, which limits the duration of wearing the device [18]. Current mechanical designs are not accurate enough to follow the gait pattern of an average human [8], but providing an effective and comfortable mechanical interface with the human body is an attainable ambition [18]. Considered exoskeletons assist only with walking, standing and with stairs, but there is also demand for devices that allow doing other tasks, such as bending over, sitting down, or entering a car etc. without restriction of movement [7] pp. 2-3, [8]. For some impairments, e.g. paraplegia, the exoskeleton requires additional crutches for walking and sit to stand movements [8]. Misalignment of the exoskeleton system is unpleasant for the user and may cause metabolic cost or pain. Therefore, the adjustability is essential to ensure that the device fits the user properly.

Personalization of the device increases the comfort for the user. All individuals are different as they have various gait disorders, so exoskeleton design should also be individualised for different users, especially in assistive home use exoskeletons [7] p. 368. Individualization improves UCD of exoskeletons since the user’s characteristics and needs can be noticed [7]. Misalignment can also be reduced by tailoring the device’s size and padding to the user, allowing both the structural design and control strategies to address specific gait disorders. Better personalization makes the device more fitting and reduces skin integrity. In addition to gait specific adjustments, physical changes in the user’s body, such as gaining or losing weight, can affect the fit and alignment of the exoskeleton. In long-term use, it is desirable to have adjustable components to accommodate such changes and prevent discomfort.

4.3 Safety

Safety, in addition to comfort, is the most considered exoskeleton factor [9]. In [9], participants raised concerns over potential harms from the technology in terms of both pressure issues and risk of falls. Due to these concerns wheelchairs remained to be the more effective and safer option.

Exoskeletons need to prevent falls, joint hyperextension, and unnatural movements. Real-time control accuracy is critical to avoid unsafe movements or falls. Sensor errors or latencies may cause mistimed or excessive actuation. Uneven terrain, obstacles, or crowded spaces increase the fall risk [13]. Stability assistance and adaptation are important for safety in uncontrolled environments. Control strategy designs need to prioritise safety without losing on functionality. Mechanical limits and joint stops are also important for reducing risk of falling but also to prevent hyperextension and unnatural movements.

In addition to comfortability, alignment is also crucial for safety. Misalignment of the exoskeleton increases injury risk; therefore, it should be minimised through good design. Right amount of pressure distribution also decreases skin injuries and pressure sores. The design features, e.g. frame shape, weight distribution and padding are important also for user safety.

One challenge for users is to trust exoskeletons, as feeling safe and being safe are different. Psychological safety is also important for trust and adoption. Knowing what the exoskeleton is doing, and when and how it will move increases confidence in using it. Fast reactivity and intuitiveness of the device also adds trust. The use of crouches provides an additional sense of control and responsibility over balance.

4.4 Ease of use

Ease of use is a critical aspect of UCD, as it directly impacts the users' willingness and ability to use the device in daily life or rehabilitation. It includes factors such as how easily the exoskeleton can be operated, adjusted, and maintained. Exoskeleton devices are for example often bulky and heavy leading to donning and doffing being difficult without assistance. Adoptability of the device can decrease due to complexity issues with wearing the device, especially in home settings where professional support may not be available. In addition to independent wearability, the ability to take off the device is an important factor which could reduce adoptability when left unaddressed. These factors are affected by device weight, attachment mechanisms, and adjustment complexity. Designing for both mechanical adjustability and ease of fitting is therefore a key UCD consideration. The possibility of PU using the device independently was listed as a nice-to-have design feature [7] p. 83. However, independent exoskeleton usage can also be restricted as for example for ReWalk, for which ease of use is lowered by the constant need for a specially trained assistant, as it can be difficult to achieve.

User interface and control system determine the intuitiveness and independence of the operation of the device. A well-designed interface reduces cognitive load and increases user confidence in managing the exoskeleton's functions. In contrast, having a complex user interface and control system can overwhelm the user, depending on the required training and cognitive demands. Exoskeleton control can be fully automated or have buttons, joysticks, or touchscreens. Ease of use can also be affected by issues causing movement to become slow and rough [9]. Therefore, use of an exoskeleton for functional daily tasks was rated lower in potential reasons for use than others [9]. Slow walking speed in daily usage can cause frustration for the user, further affecting the user's adaptability with the device. User experience is improved by comfort and safety, which leads to better meeting of UCD requirements.

Exoskeletons require time and effort for a user to learn how to operate the exoskeleton safely and effectively. For example, HAL-5 exoskeleton required two months to calibrate it optimally for users [12]. This highlights the importance of individual adaptation and system tuning. In home use, the exoskeleton can be calibrated for the user optimally [12], which also supports comfort, safety and effectiveness. Synchronizing the movement of the exoskeleton with the user's gait plays a major part in this process. Synchronization reduces the user's adaptation time and physical strain during use [26]. Poor synchronization can cause delays, misalignment, or discomfort, which can weaken safety and the user's trust in the device. Minimisation of the learning curve and the optimisation of calibration and synchronisation processes are important for increasing user confidence, independence, and sustained engagement. A great demand for training can decrease the trust and confidence toward exoskeleton use, which can impact usability and adoption.

Maintenance and daily usability address the ease of exoskeleton management in everyday situations including tasks like charging, cleaning, and basic adjustments of the device. Everyday use also places additional demands on usability and durability, as the device must function reliably in various home use settings [22]. Furthermore, having access to service centres or technical support may be difficult while traveling or if the user lives in a remote area. This highlights the need for reliable, easy maintenance devices, as lack of maintenance can lead to longer downtimes and reduced usability. Ease of use is a fundamental factor in improving user acceptance, confidence, and consistent engagement with the exoskeleton.

4.5 Cognitive and Social aspect

Cognitive and social demands, and various factors influence exoskeleton design and adaption in addition to the physical and functional considerations. These factors can play a major role in user experience and integration of the device into daily life. The exoskeleton use is impacted by the user's mental workload, user confidence, social perception, and interaction with others.

Cognitive aspects are important to remember while designing exoskeletons. In cognitive terms, the user's trust in the device needs to reach a level where the device is safe to walk with, as the certified and perceived safety are not the same [1]. Perceived safety means the user's perception of the level of danger when interacting with an exoskeleton [1]. Meaning that user may perceive an exoskeleton as unsafe or scary, even if it is a certified robot and considered safe objectively. For some users, their muscles activate without the user detecting if it is a part of the normal exoskeleton usage, or not [1]. This can also add to the unsafe feeling for the users and increase fear of falling. Fear of falling can be mentally draining and contributes to cognitive load. Cognitive load refers to the mental effort required from the user to operate the exoskeleton while managing other tasks and environmental demands. Complex controls, multitasking, and the need for constant attention increases the user's cognitive demands. Daily living often requires using hands while standing [7] pp. 2-3, such as picking items from a shelf, which adds additional challenge. People with gait impairments need to focus intensely on walking with the exoskeleton, so additional tasks increase the load. UCD aims to minimize these unnecessary cognitive demands by creating intuitive and supportive control systems [1]. Reducing cognitive load also improves usability, comfort and user confidence.

Social aspect considers the user's perception of other people's views and the interaction with them. Beyond physical support, exoskeletons can influence how users engage socially and emotionally with their environment. In the survey [9], eye-level social interaction, and the confidence, hope, and joy standing and walking could provide where recognised by many users. Consequently, standing and walking can benefit also psychosocially. The ability to stand and walk can increase the feeling of belonging with others, as the exoskeleton user interacts with the environment in a similar way. However, the user may also feel excluded socially, as the exoskeleton can draw unwanted attention, and the user can become self-conscious in public, potentially contributing to social isolation [1].

In a survey of prioritised design requirements for potential users of a soft assistive exoskeleton [7] pp. 83-84, the users rated "Worn under users' usual clothing" and "Compatible with users' preferred clothing" as must-have design features. Although, in a wheelchair user survey [9], the appearance of the exoskeleton was rated as a "neither important nor unimportant" feature. The difference in how appearance is valued between user groups can be explained by their differing levels of mobility. Potential soft exoskeleton users are often able to walk independently to some extent, and therefore may prefer to conceal both their impairment and the device. In contrast, wheelchair users who cannot stand or walk unaided, prioritize the ability to walk over the appearance of exoskeletons. The social experiences affect the exoskeleton use and cognitive load. For instance, if the user feels judged, it increases anxiety and reduces user comfort, which increases mental effort to perform tasks, whereas positive experiences can have opposite effects.

Assistive exoskeletons can improve user independence which can change the dynamics between the PU and SU. Exoskeleton use may shift the roles with caregiving relationships, as the user can act more independently or may require new forms of assistance. This affects more home settings with family members, as the routines change. For example, a wheelchair user may not require constant help with reaching things from shelves as they could stand for themselves.

5 Conclusions

This thesis identified the most critical technical and user-centred features affecting lower-limb exoskeletons usability, highlighting the importance of user-centred design. Exoskeletons can improve independence and quality of life for patients with walking impairments. Exoskeletons enable repetitive motor movements that stimulate the brain to reconstruct neural pathways which enhance movement rehabilitation. However, the technology still needs to advance in order to enable more widespread daily use exoskeletons for functional purposes. The design must follow the biomechanics of the user's natural gait pattern and simultaneously ensure that exoskeletons are user-friendly focusing on comfort, safety and ease of use.

While UCD highlights the importance of meeting user needs, the development of exoskeletons inevitably involves balancing competing design priorities. Technical, mechanical, and ergonomic factors frequently interact in complex ways, requiring trade-offs in designing. For example, increasing stability and durability increases weight or rigidity of the exoskeleton's frame, which can simultaneously reduce mobility and user comfort. Similarly, adding more actuators and sensors may improve control precision, but at the cost of greater complexity, higher maintenance demands, and additional cognitive load for the user. Designers must navigate these trade-offs to create device that achieve sufficient support while preserving freedom of movement, adaptability, and long-term wearability.

In the end, exoskeleton development is a process of balancing safety, functionality, comfort, and usability, while recognizing that improvements in one area may impose limitations in another. The main considerations in exoskeleton development identified in this thesis are safety, functionality, comfort, and usability. It needs to also be recognised that improvements in these areas may impose limitations in another, leading the development process to be one of balancing features and goals. In addition, the UCD requirements depend on the goal of the user and the type of the exoskeleton.

At the current state, exoskeletons encounter numerous UCD-related challenges. The users may consider conventional assistive and rehabilitative devices more trustworthy and accessible compared to exoskeletons. Additionally they are too expensive for assistive home use. In the future, exoskeletons have the potential to reduce care costs and thus support economic benefits, but their cost must first decrease before achieving this goal.

Although the number of studies discussing UCD considerations, device structures, device components, and physical impairment causing actors is large and has grown significantly in the recent years, papers concerning overground exoskeletons and their designs were found to be limited. Overground exoskeletons are relatively new technology and are not yet widely used for assistive purposes. In addition, only a few studies were empirical introducing a need for further primary data from PUs, SUs, and TUs.

Another limitation of this thesis is that the study excluded industrial and military applications of exoskeletons. This was done, as the exclusion allowed for a more focused and comprehensive study into rehabilitative and assistive use. Industrial and military uses often help advance technologies and as such are viable considerations for future research.

Mechatronical development of exoskeletons, although more dependant on technological advancements, should be researched in order to reduce the weight or price of exoskeletons. Lowering the price of exoskeletons could lead to more widespread use of such devices. The question still remains if exoskeletons could be used by the general public in the future, as current technology is not sufficient enough to allow for assistive home use strongly due to price issues. Rehabilitative exoskeletons are less demanding than assistive devices, making them more accessible, although not guaranteed, in clinical settings, however they also experience similar issues especially with high prices and user interface.

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