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Compensating impaired movements: design principles for lower-limb exoskeletons

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Abstract

Walking is one of the top priorities of patients that suffer from reduced locomotor function. Exoskeletons bear the promise of restoring that function by providing active support, and thus, enable economical movements again. However, there is a disparity of support strategies and designs with limited clinical validation. The fundamental requirements that drive these designs are manifold and partially lack validated approaches. The purpose of the thesis is to propose and validate essential requirements for assistive exoskeletons independent of their technical implementation while also ensuring causal tracebility. The requirements have been derived from mechanisms and features of efficient movements and prevalent design principles of devices that are able to improve movement economy. As characteristics of unimpaired movements do not necessarily apply to patients, sources of pathological movements and design principles of movement aids have been identified and compared. To determine the most beneficial support method, an assessment framework based on gait deviations and underlying causes has been proposed. The requirements have been implemented in a first prototype: the Myosuit. The Myosuit was used for initial validation tests on a single participant during sitting transfers. The requirements have been further validated with regard to their feasibility and usability in five patients with different pathologies.

The assessment framework based on gait deviations identified external support of the knee extensor muscles and flexor muscles of the hip, knee and ankle joint as the most promising interventions to improve impaired walking. When considering other movements besides walking, also supporting hip extension can provide a benefit. A set of essential requirements and design principles related to the exoskeleton interface, provision of force and controls have been derived and implemented in the Myosuit. The Myosuit uses a bi-articular, cable driven design to assist knee and hip extension simultaneously and a posture based control approach. Elastic elements support the flexor muscles antagonistic to the active cable drive. The Myosuit was able to reduce extensor muscle activity during sitting transfers and could be used without any serious adverse events in a patient population during a feasibility test. These initial tests validated the essential requirements and related design principles. This shows that the applied methods of deriving the requirements can lead to an exoskeleton design that fulfill basic needs of various pathologies and may provide a functional benefit.

Überblick

Das Gehen ist eine der obersten Prioritäten von Patienten, die unter eingeschränkter Bewegungsfunktion leiden. Exoskelette versprechen diese Funktion durch aktive Unterstützung wiederherzustellen und ermöglichen so wieder effiziente Bewegungen. Allerdings gibt es eine Vielzahl von Unterstützungsstrategien und Designs mit begrenzter klinischer Validierung. Die Anforderungen, die diesen Designs zugrunde liegen, sind vielfältig und es fehlen häufig substantielle Ansätze. Ziel der Arbeit ist es, wesentliche Anforderungen an unterstützende Exoskelette unabhängig von ihrer technischen Umsetzung abzuleiten und zu validieren, wobei auch die kausale Nachverfolgbarkeit sichergestellt wird. Die Ableitung der Anforderungen erfolgte durch die Identifizierung von Mechanismen und Funktionen, die effiziente Bewegungen ermöglichen, und vorherrschende Konstruktionsprinzipien von Geräten, die die Bewegungsökonomie verbessern können. Da diese Merkmale nicht unbedingt auf Patienten zutreffen, wurden die Ursachen pathologischer Bewegungen und die Konstruktionsprinzipien von Bewegungshilfen identifiziert und verglichen. Um die vorteilhafteste Unterstützungsmethode zu identifizieren, wurde ein Bewertungsrahmen entwickelt, der auf Gangabweichungen und den zugrunde liegenden Ursachen basiert. Die Anforderungen wurden in einem ersten Prototyp umgesetzt: dem Myosuit. Der Myosuit diente als Mittel für erste Validierungstests an einem einzelnen Probanden während mehrfacher Aufsteh- und Hinsetzvorgänge. Die Anforderungen wurden ausserdem im Hinblick auf ihre Machbarkeit und Anwendbarkeit an fünf Patienten mit verschiedenen Pathologien validiert.
Im Rahmen der Suche nach der optimalen Unterstützungsmethode wurde die externe Unterstützung der Kniestreckmuskeln und der Beugemuskeln des Hüft-, Knie- und Sprunggelenks als die vielversprechendsten Interventionen zur Verbesserung von Gehbehinderungen identifiziert. Wenn neben dem Gehen auch andere Bewegungen berücksichtigt werden, kann auch die Unterstützung der Hüftstreckung einen Nutzen bringen. Grundlegende Anforderungen in Bezug auf die Mensch-Maschinen-Schnittstelle, die Bereitstellung der Unterstützung und die Steuerung wurden abgeleitet und im Myosuit umgesetzt. Der Myosuit verwendet ein bi-artikuläres, kabelgesteuertes Design zur gleichzeitigen Unterstützung der Knie- und Hüftstecker unter Verwendung eines haltungsbasierten Steuerungsansatzes. Elastische Elemente unterstützen die Beugemuskeln antagonistisch zum aktiven Kabelantrieb. Der Myosuit war in der Lage, die Streckmuskelaktivität zu reduzieren und konnte während des Machbarkeitstests ohne schwerwiegende unerwünschte Ereignisse in einer Patientenpopulation eingesetzt werden. Diese ersten Tests validierten die wesentlichen Anforderungen und die damit verbundenen Konstruktionsprinzipien. Dies zeigt, dass die angewandten Methoden zur Herleitung der Anforderungen zu einem Exoskelett-Design führen können, das grundlegende Bedürfnisse verschiedener Pathologien erfüllt und einen funktionellen Nutzen bieten kann.

Acknowledgements

Throughout the writing of this dissertation I have received a great deal of support and assistance.

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Symbols

Symbols

EC	encoder counts [1]
F	force [N]
g	gravitational acceleration $[m/s^2]$
h	height [m]
k	spring stiffness [N/m]
l	length [m]
m	mass [kg]
m_{force}	suit stiffness [N/rad]
m_{hip}	hip-encoder mapping [deg/rad]
m_{knee}	knee-encoder mapping [deg/rad] [kg]
M	moment [Nm]
P	power [Watt]
$r_{joint\_ligament}$	moment arm ligament [m]
v	velocity [m/s]
W	work [Joule]
α	absolute angle [deg]
β	relative angle [deg]
θ	acceleration $[m/s^2]$
ω	angular velocity [rad/s]

Acronyms and Abbreviations

AFO	ankle foot orthosis
ADL	activities of daily living
BMI	body mass index
COM	center of mass
CT	contracture
DF	dorsi-flexion
EC	encoder counts
EXT	extension
FES	functional elctrical stimulation
FIX-N	fixed in a neutral position
FLE	flexion
IMU	inertial measurement unit
iSCI	incomplete spinal cord injury
KE	kinematic energy
MCID	minimal clinically important difference
MOCA	Montreal Cognitive Assessment test
MS	Multiple Sclerosis
NRS	Numeric Rating Scale
PE	potential energy
PF	plantar-flexion
PT	physiotherapist
ROS	roll-over shape
RQ	requirement
SD	standard deviation
SP	spasticity
SUS	system usability scale
WEIMuS	Würzburger Erschöpfungs-Inventar bei MS
WK	weakness

Contents

Preface

The work presented in this thesis was conducted at the Sensory-Motor Systems Lab at ETH Zurich in collaboration with MyoSwiss AG. I co-founded MyoSwiss AG as ETH Spin-off resulting from the positive results presented in chapter 5 and the patent application:

K. Schmidt, J.E. Duarte, R. Riener, "Apparatus for supporting a limb of a user against gravity", EP3342390, priority date 29.12.2016

Since the foundation of MyoSwiss in 2017, I act as Chief Technology Officer and board member, and since 2020, I act additionally as Chief Executive Officer for the European subsidiary based in Germany. The continued development of the Myosuit towards certification as medical device and related intellectual property was completely managed by MyoSwiss AG.

Parts of the introduction and chapter 5 are based on the following scientific article that has been published:

K. Schmidt K, J.E. Duarte, M. Grimmer, A. Sancho-Puchades, H. Wei, C.S. Easthope, R. Riener, "The Myosuit: Bi-articular Anti-gravity Exosuit That Reduces Hip Extensor Activity in Sitting Transfers", Frontiers in Neurorobotics, Oct 2017

The scientific investigations reported in chapter 6 were conducted in close collaboration with Dr. Martin Grimmer. Parts of the introduction and chapter 6 are based on a publication with a shared first-authorship:

M. Grimmer, K. Schmidt K, J.E. Duarte, L. Neuner, K. Koginiv, R. Riener, "Stance and Swing Detection Based on the Angular Velocity of Lower Limb Segments During Walking", Frontiers in Neurorobotics, vol. 13, 2019

The feasibility test presented in chapter 7 was conducted in collaboration with licensed physiotherapist Nathalie Zwickl as part of her thesis to obtain the title Master of Science in Physiotherapy.

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Chapter 1

Introduction

1.1 Impaired movements

Mobility and independence are key determinants for quality of life [5, 6]. When mobility is limited, a person's quality of life can be impacted negatively. The main causes for mobility limitations are reductions in physical capacity with increasing age [7], diseases, or injuries. From the age of 60 to the age of 85, the mean steps per day decrease by about 77% [8]. This can negatively affect a person since a higher number of steps per day is associated with several positive health outcomes including reductions in body mass index [9], risk of cardiovascular disease [10], and mortality [11]. Along with aging, neurological injuries such as spinal cord injury (SCI) can also limit a person's mobility. For example, incomplete SCI patients (ASIA grades C and D; 30% of SCI patients) have full range of motion and the ability to move against gravity with at least half of the key muscles. However, walking and standing abilities, which are top priorities for this patient group [12], are clearly limited for most of the patients [13].

While exercise can help mitigate the reductions in strength [14] and stamina [15], the overall trend cannot be stopped [15, 16]. To be mobile and move independently, the best option for many people is the use of technologies that can compensate their impairment. These technologies range from passive devices, such as orthoses or wheelchairs, to powered devices, such as exoskeletons. Especially the wheelchair is an efficient tool to maintain mobility. But in terms of versatility and health related impact it cannot compete with bipedal locomotion. This might be the main reason why many patients report that recovering walking capacity is one of their top priorities [17, 18, 19, 20, 21, 22]. There is a great need for devices that can support and facilitate the rehabilitation process and help patients to reach their potential rapidly. But even if the rehabilitation potential is exhausted and patients only improve incrementally, many still require assistance and support of locomotion tasks during their everyday life. In these cases, their most promising option to restore mobility and independence might be robotic exoskeletons.

1.2 Robotic exoskeletons

Given the great need for devices that can restore function related to specific movements, it is not surprising that there has been a multitude of exoskeleton developments. A recent systematic review identified 25 lower-limb exoskeletons that are intended for gait training in neuromusclular impairments [23]. Only six out of these 25 devices are commercially available. Yet two other systematic reviews found a total of 239 additional exoskeletons for the lower-limbs [24, 25]. Considering the high number of developments, it is surprising that exoskeletons are still not part of the daily life of patients. There seems to be a mismatch between the numerous developments and what is available and suitable for patients. Despite all the progress that has been made, exoskeletons are also still not recommended by official rehabilitation guidelines [26, 27]. There is a clear lack of evidence that these devices are are superior to conventional interventions. The great number of designs and the fact that these devices are not commonly used outside of research indicates that the requirements utilized to design these devices are insufficient and do not meet the user's needs.

1.2.1 Compensation of impaired movements

Exoskeletons have the potential to help patients to achieve their goals within therapy and related to upright mobility. When exoskeletons are designed to fully drive the kinematic pattern of a user, controllers and actuators can rely on trajectories of unimpaired movements [28]. This approach is usually implemented in exoskeletons that are intended for paralysed or very weak patients that could not move without the device. However, design decisions become more complicated when exoskeletons are designed to only partially support, and thus, provide assistance for residual functions. It can often be observed that patients with the same pathology move very differently. Patients usually have their own unique set of deficits that result in different manifestations of movement patterns. They substitute their lost functionality by using various compensatory movements. These individual characteristics make it difficult to develop general approaches that can provide a benefit for many patients. This exacerbates the design process considerably. However, there seem to be general approaches that can support various pathologies. One such an approach is reflected in crutches. They provide partial weight-bearing that benefits a wide range of patients independent of their pathology. Similarly, patients seem to profit from support of toe-clearance to swing the leg forward by using ankle-foot orthoses (AFOs). These devices are recommended by the guidelines because they are simple, light-weight and efficient [26, 27, 29]. Exoskeletons could be also designed for basic functions while keeping their weight and level of complexity low. Nonetheless, it is not obvious which deficits exoskeletons should target and where they can be advantageous to conventional solutions.

1.2.2 Essential requirements for exoskeleton designs

The complexity due to the uniqueness of deficits and resulting movements require a structured design approach. Design requirements need to be derived from the targeted movements while also considering how deficits prevent their execution. Based on these insights, general support strategies could be deducted. Further requirements would also need to be elicited from the various exoskeleton designs that already provide data about their positive impact on movement economy. Such an approach would ensure that the device's design targets relevant deficits that could provide a benefit to a multitude of patients, and thus, could serve for new developments. To be used as a basis for new developments, these requirements need be independent on any technical implementation. That is why they are called essential requirements within the scope of this thesis. These requirements can potentially be used to derive design principles for implementations of exoskeletons that assist residual function. Simultaneously, the essential requirements could also be used as a framework during design validations. By following this approach, it becomes more likely that patients receive support that compensates their impairment which results in improved movement function.

1.3 Thesis goal and outline

The goal of the thesis is to find and validate general design approaches for exoskeletons which are reflected in essential requirements and design principles. They can serve for new developments of exoskeletons that are intended to support residual functions. This thesis presents a structured approach to derive these essential requirements. Chapter 2 identifies mechanisms and features that make walking efficient and devices that can improve movement economy. As pathological movements are likely to hinder these mechanisms, the underlying causes are systematically analysed and compared to existing devices that compensate such impairments in chapter 3. Due to the disparity of support strategies, this chapter develops a framework that allows to determine a support strategy that directly addresses the sources that hinder efficient and economical movements. The framework focuses on deviations and underlying causes in pathological gait. Based on the discussion of the essential requirements, additional design principles are derived in chapter 4 and implemented in the the design of the Myosuit in chapter 5. The Myosuit is an exoskeleton for partial support that combines active and passive actuation methods to enable the device to be light-weight and non-obtrusive while providing considerable assistance to the user. To use such a system during cyclic movements like walking, suitable phase segmentation methods are presented in chapter 6. Chapter 7 presents a feasibility test of the Myosuit with five patients. The feasibility test is used to validate the essential requirements. Eventually, the results of the validation are discussed in chapter 8.

Chapter 2

Normal Gait

To develop strategies to support impaired movements it is necessary to identify features that enable efficient and economical locomotion in unimpaired movements. People with limited mobility frequently report that regaining walking capacity is one of their top priorities [17, 18, 19, 20, 21, 22]. That is the reason why this chapter focuses on walking, and specifically on normal gait.

The distinct characteristics of normal gait and the associated functions of the biological joints and muscles are introduced in section 2.1. After establishing the basics and terminology of normal gait, this chapter focuses on the identification of mechanisms and features that enable an efficient and economical movement. That is why section 2.2 introduces three theories related to efficiency of normal gait. The theories include the six determinants of normal gait, dynamic walking and energy transfers in multi-segment systems. The derived insights serve as a basis to answer the question posed in section 2.3: What makes walking efficient? To find out how these mechanisms and features can be exploited in normal gait, section 2.4 presents assistive devices that are able to reduce the energy demand during normal walking beyond the energy demand without the device. The conclusions from the presented theories related to walking efficiency and the properties of these devices were used to derive the first essential requirements in section 2.5. These requirements include generic approaches that show high potential to be beneficial in assistive exoskeletons that aim to compensate impaired movements.

2.1 Basics of normal gait

2.1.1 Bones, joints and muscles of the lower limbs

This section establishes the terminology related to bones, joints and muscles of the lower limbs that are referred to in subsequent parts of the thesis. Figure 2.1 depicts the joints and main bones of the lower limbs. At the proximal part, the pelvis connects with the lumbar spine. Even though the pelvis consists of multiple bones that allow small relative movements, in gait analysis it is considered to be a single rigid structure. At its lower part, it composes the socket for the head of the femur creating the hip joint. The femur extends distally until the bone widens and forms a groove that articulates the patella and connecting to the tibia constituting the knee joint. The tibia articulates with the femur at its proximal end and provides the upper and medial surfaces of the ankle joint. On the lateral side of the tibia, there is the fibula which broadened lower end forms the lateral part of the ankle joint.

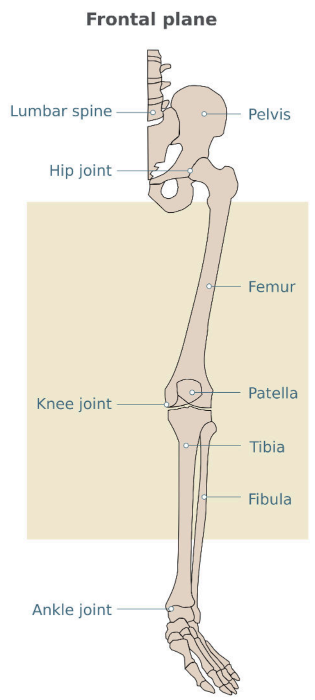


Figure 2.1: Joints and bones of the lower limbs.

Figure 2.2 shows the muscles of the lower limbs. Gluteus maximus (extensor) and illiopsoas (flexor) are acting antagonistically across the hip. The quadriceps (vastus medialis, vastus intermedius, vastus lateralis, rectus femoris) are the main knee extensors including one bi-articular muscle (rectus femoris) that is also a hip flexor. Antagonistic to the quadriceps, the hamstrings act both as knee flexors and hip extensors. Gastrocnemius and plantaris are knee flexors as well. At the same time, these bi-articular muscles are also ankle plantar-flexors. Plantar-flexion is additionally supported by soleus which only spans the ankle joint. Antagonistic to the plantar-flexors, tibealis anterior is acting as dorsi-flexor.

2.1.2 Tasks, phases and periods of normal gait

Normal gait refers to the absence of pathological movement patterns that result in increased required energy expenditure. It consists of repetitive events and features that are optimized to ensure ambulation while conserving energy throughout the movement. Due to the repetitiveness of the events and features, normal gait can be divided into distinct phases, and thus, it can be described as a repetitive cycle. Each gait cycle is defined as the interval between two successive instances observed on one leg at which the foot contacts the ground [1].

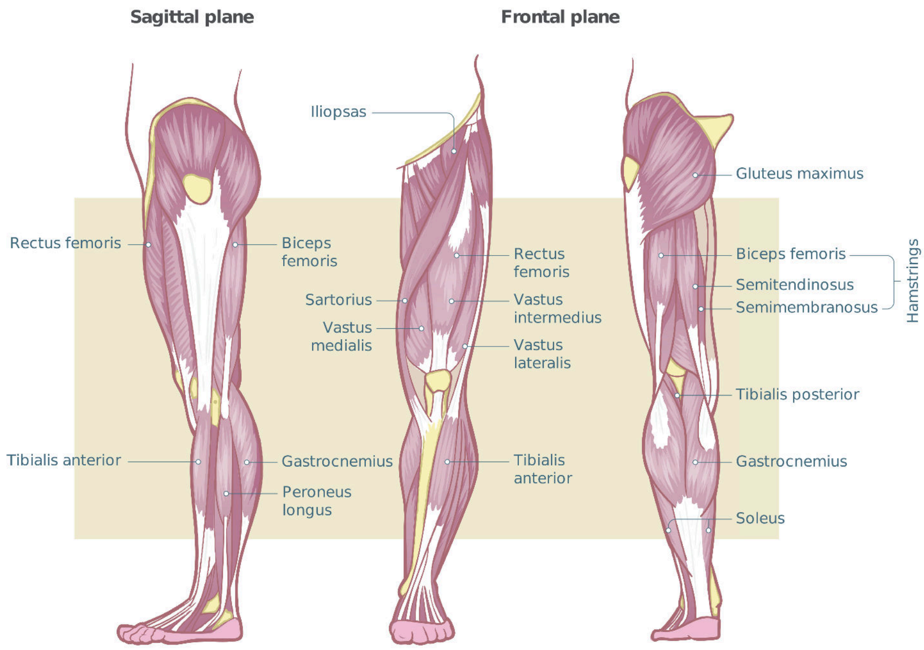


Figure 2.2: Main muscles and muscle groups of the lower limbs.

The time shift between the cycles of the opposite legs is depicted in figure 2.3. After a foot contacts the ground there is always an interval of double support. During double support, both feet are on the ground to allow the weight shift from one leg to another. One leg is leading meaning that the foot is placed anteriorly. The other leg is trailing with an posterior foot placement. The leading leg just finished swinging forward and the trailing leg finished single leg support. Single leg support allows the contralateral limb to swing forward while maintaining upright posture. Each cycle contains two intervals of single leg and double support while maintaining a constant time shift between the events at the same speed [1].
Figure 2.3 shows that one gait cycle equals one stride. One stride can be divided in different phases and periods by certain events. These events are:

- Initial contact
- Opposite toe off
- Heel rise
- Opposite initial contact
- Toe-off
- Feet adjacent
- Vertical tibia

One gait cycle consists of two phases, namely stance and swing phase. Initial contact and toe-off mark the transition between the two. Stance phase can be characterized by two main tasks: Weight acceptance and single limb support. The leg that started a new cycle and just touched the ground must bear the person's weight without collapsing while maintaining static and dynamic balance. The main task of swing phase is limb advancement. Stance phase lasts about 60% of the cycle and swing phase for the remaining 40%.
The periods within stance phase are:

- Loading response
- Mid-stance
- Terminal stance
- Pre-swing

Swing phase is subdivided in:

- Initial swing
- Mid-swing
- Terminal swing

Each period can be described by the orchestration of muscles to ensure continuous and symmetric movements. The following two sections (2.1.3 and 2.1.4) describe the the periods and role of muscle groups in more detail.

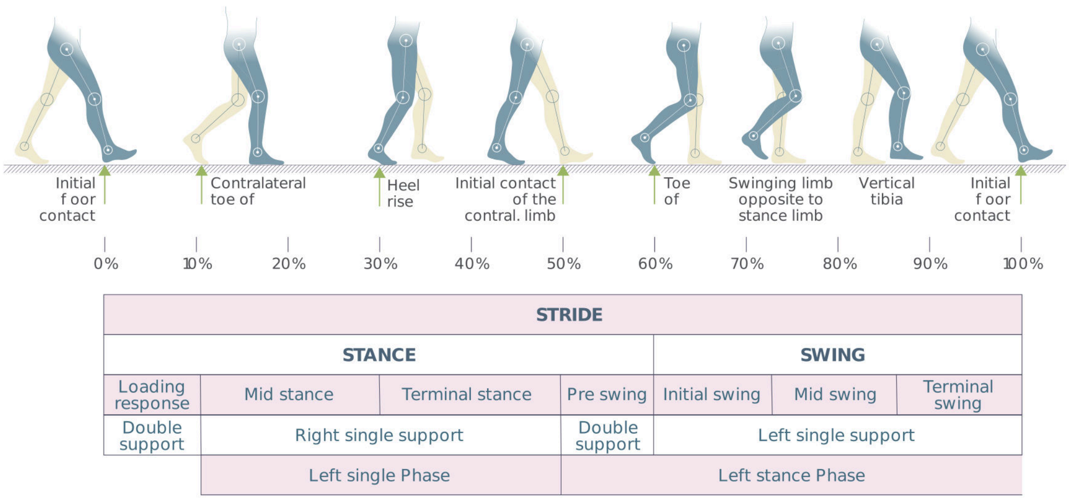


Figure 2.3: Phases, periods and tasks of normal gait.

2.1.3 Stance phase

Initial contact decelerates the body as the foot reaches the floor. This is achieved by simultaneous activity of the knee extensor and flexor muscles. The concurrent activation of antagonistic muscles stabilizes the knee and ensures that the subsequent loading of the leg does not collapse it. The hip extensors also contract to decelerate the thigh and to support knee extension in anticipation of the loading of the leg. The tibialis anterior begins a lengthening contraction (eccentric contraction) at the same time preventing the foot from slapping down on the floor. During loading response, the knee extensors contract and the leg accepts the body weight. The knee bends slightly and begins to extend under the shortening contraction (concentric contraction) of the knee extensors to soften the initial impact. This extension is supported by a lengthening contraction of the plantar-flexors. They move the contact point of the foot forward and shift the reaction force vector of the body anterior of the knee aiding with knee extension. At the same time, the gluteus medius contracts isometrically to stabilize the pelvis in the frontal plane. The center of gravity has reached its highest point in mid-stance. This potential energy is carried forward by momentum with little energy. While the knee remains extended with the foot on the floor the body weight falls forward. This transmission of energy is aligned with the direction of progression during normal gait. The lengthening contraction of the soleus muscle keeps the forefoot pressed against the floor. The created force couple linkage supports knee extension without requiring muscle activation of the quadriceps. The knee stays passively extended in terminal stance. That is due to the force coupling created by knee extension and ankle plantar-flexion that continues to support the extended knee. Other than in the previous period, the ankle plantar-flexors begin a shortening contraction to accelerate the body forward generating a burst of energy. The created burst is responsible for most of the power generation during normal gait and propulses the body forward. In preparation of the unloading with the aim of swinging the leg forward, the hip flexors also become active. The opposite leg begins to accept weight in pre-swing which allows the leg to prepare for its own swing phase. The ankle plantar-flexors are no longer active and the hip flexors begin to lift the leg in preparation of swinging forward. This is achieved by shortening contractions of the hip flexors. Since the leg behaves predominantly as a passive pendulum in the subsequent swing phase, energy is only required for brief muscle activity of the hip flexors.

2.1.4 Swing phase

The passive pendulum characteristics affect the duration of swing and the length of the stride. There is a direct dependency on the length of the freely swinging leg and the dynamic friction of the knee joint and the associated tissue. Any prolongation of hip flexor activity or premature activity of the hamstrings changes the geometry of the swinging leg, and thus, affects swing duration and stride length. The hip flexors stop being active in initial swing as the leg is using the passive pendulum characteristics to swing forward. At the same time, the ankle dorsi-flexor begins a shorting contraction to allow the foot to clear the floor.

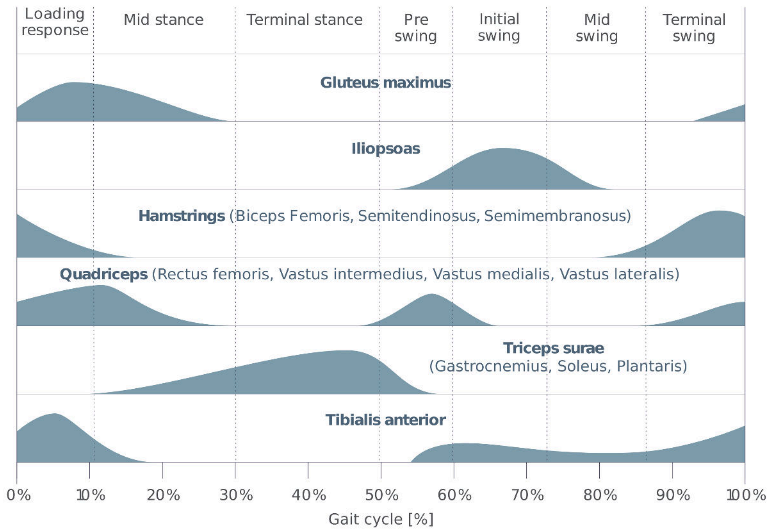


Figure 2.4: Muscle activation during normal gait (data from [1]).

The passive pendulum action of the leg and the dorsi-flexor activity for foot clearance continues during mid-swing until terminal swing. The leg begins to actively decelerate the swinging leg by contracting the hamstrings eccentrically or isometrically in terminal swing. This slows hip flexion and knee extension efficiently. At the same time, early quadriceps activity prepares the knee for weight acceptance. The tibialis anterior shows an isometric or lengthening contraction to ensure the foot is kept at a slightly dorsi-flexed position before touching the ground again. Detailed muscle activity patterns can be found in Figure 2.4.

2.2 Theories related to the efficiency of normal gait

This section introduces three theories related to the efficiency of normal gait. Insights from these theories are used in section 2.3 to summarize what mechanisms and features make walking an efficient task.

2.2.1 Six determinants of normal gait

It is hypothesized that the human body orchestrates motions of various segments and controls the activity of different muscles so that the metabolic energy required for a given task is minimized [3]. A theory first introduced by Saunders et al. described six determinants that make gait an efficient task [30, 31]. It is assumed that there are certain kinematic features that reduce the displacement of the body's center of mass (COM) by transferring intersecting arcs to sinusoidal movements which in turn also reduces energy expenditure. This theory describes the determinants as

1. pelvic rotation,

2. pelvic obliquity (pelvic drop),

3. knee flexion during early stance,

4. ankle mechanism,

5. foot mechanism, and

6. lateral displacement of the body (hip abduction).

The determinants of gait have been generally accepted in the past without any empirical data or objective evaluation provided [32, 33, 34, 3]. However, more recent publications suggest that some of these kinematic features may not affect gait in the originally hypothesized way [35, 36, 37, 38, 39].

Pelvic rotation refers to to rotation of the pelvis around a vertical axis that brings the hip forwards during hip flexion and backwards during hip extension. The rotation reduces the required flexion and extension range at the hip for a given stride length. This reduction, in turn, results in a decreased vertical movement of the hip. Pelvic obliquity reduces the vertical trunk movement by tilting the pelvis about an anteroposterior axis. When the hip of the stance phase leg is at its highest point, the pelvis drops downwards on the contralateral side. This reduces the average height of the hip joint and decreases the vertical execution of the trunk. Knee flexion during early stance adjusts the effective leg length so that the hip joint does not rise as high if the knee would be kept extended. However, more recent studies suggest that the first three determinants do not reduce the vertical displacement during normal gait [35, 36, 38]. The second and third determinants are likely to support shock absorption that is caused by the discontinuities introduced by heel strike [37, 39]. The fourth and fifth determinants concern the foot and ankle complex and it is hypothesized that the related mechanisms create a nearly circular rocker [40]. At heel strike, the heel sticks out behind the ankle joint and lengthens the leg effectively. The forefoot lengthens it in a similar way at the end of stance phase. This behavior is also reflected in the center of pressure between the foot and the ground as it follows a circular trajectory. The kinematic features related to the foot and ankle complex are the only ones that appear to actually flatten the COM trajectories [3].

Even though the theory of the six determinants suggest that a minimization in vertical displacement is one of the main drivers to reduce energy expenditure, the human body does not seem to optimize the COM trajectory to approximate a flat line parallel to the direction of progression. It has been shown that energy expenditure increases by a factor of two or more when humans intentionally try to reduce vertical displacement of the COM while walking or when taking shorter steps [41, 42]. The human body exploits energy transfers from potential to kinetic energy. Vertical displacements are not necessarily inefficient and do not increase energy expenditure if the movement can be used for storing and returning mechanical energy. Nonetheless, this can only work sufficiently well if there are efficient transfers which could be enabled by certain kinematic features as described by the six determinants of gait. Even though the six determinants are kinematic features that can be observed during normal gait, they are not necessarily the basic principles that make normal gait efficient.

2.2.2 Dynamic walking

Whereas the six determinants focus on certain kinematic features to reduce energy expenditure, the inverted pendulum model explains walking efficiency through energy conversation. It suggests that it is energetically more beneficial for the leg to act like a pendulum carrying the trunk and adhere to the trajectory of an arc [43, 44]. There is a constant energy exchange between potential and kinetic energy requiring zero work for the COM to travel along the arc defined by the leg length. It is also assumed that the swing leg acts like a pendulum (non-inverted) when moving forward in line with the direction of progression [45]. Conserving energy by exchanging mechanical energy requires no work. Consequently, there is no additional force needed to cause movement. This theory proposes an explanation why normal gait should not require any metabolic energy since the muscles don't have to perform work. This, however, contradicts anyone's experience that muscles have to contract during walking to cause a displacement of the COM. Even though the pendulum theory provides an analogy and indication how normal gait can be economical, the theory is incomplete.

McGeer developed an extension of the initial pendulum theory by including the heel-strike collision in the model. The theory is referred to as dynamic walking. It shows that the heel-strike collision can produce conditions for periodic gait without active control and minimal energy input [46]. Dynamic walking describes locomotion generated mainly by passive dynamics of the legs independent of any active powering or control. The original model was developed for two-legged machines that needed neither active control or energy input. Kuo et al. used this theory to explain the processes behind economical human gait and its clinical implications [47, 48].

This theory acknowledges that the single support leg behaves as an inverted pendulum to transport the COM with little energy. Although some metabolic energy is needed to activate the muscles ensuring that the leg is kept straight during the movement. Dynamic walking considers step-to-step transitions in addition to the initial pendulum model. These transitions require energy because the COM velocity is redirected during double support phase. It is important to point out that the COM requires redirection and not propulsion or lifting. It can be best visualized when observing the COM velocity at the end of the arc when it is approximately perpendicular to the trailing leg. The associated motion is forward and downward. To continue moving the COM in the direction of progression without falling, it requires redirection to the new arc spanned by to leading limp as shown in figure 2.5. This will result in an upward movement of the COM.

The redirection can be modeled as collision that dissipates energy. Since the leading leg is opposing the downwards path dictated by the trailing leg, the leading limb is performing negative work on the COM. Positive work is necessary to compensate for the dissipation. The work required can be performed by either ankle push-off or powering the hip [48]. Donelan et al. have shown that step-to-step transitions may require $60 - 70\%$

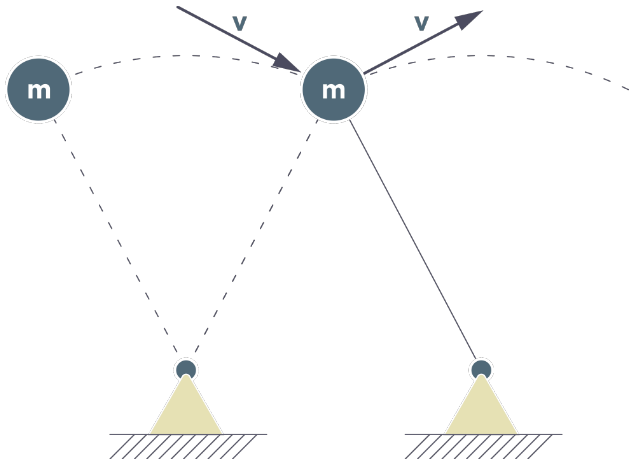


Figure 2.5: Redirection of the COM modeled as an inverted pendulum.

of overall metabolic energy for normal gait [49]. Grabowski et al. found a lower percentage, i.e. 45%, for the metabolic cost of work performed on COM redirection . This is still a considerable portion especially when compared to the 28% of total metabolic cost that is required to generate the force to support the body weight [50].

A big part of of the remaining portion of required energy cost is believed to stem from forced leg motion. The dynamic walking model indicates that forced leg motion affects step-to-step transition cost significantly. Longer or wider steps increase step-to-step transition cost. This can be explained by increased COM velocity and greater directional change during redirection. The logical conclusion would be to use narrower, shorter and faster steps to mitigate the cost of step-to-step transitions. However, this is only a trade-off since work needs to be performed to move the leg relative the body. Even though shorter and faster steps reduce step-to-step transition cost, they increase the cost of moving the legs back and forth. There seems to be a trade-off between the two which also explains why humans increase step length and frequency proportionally [51].

Two of the six determinants of normal gait are also mentioned in the dynamic walking theory. The foot and ankle complex can reduce step-to-step transition cost when it is used to act like sections of a wheel. Dynamic walking models use arc-shaped feet to reduce the negative collision work [46]. During human normal gait, the feet seem to mimic this behavior as the center of pressure is progressing on the ground like a wheel. This mechanism has been defined geometrically and is referred to as roll-over shapes (ROS) [40].

The dynamic walking model can also provide insights regarding stability during normal gait. Walking is passively stabilized with no need for direct control over the leg motions. Small perturbations can be dissipated over subsequent steps due to the energy dissipation during heel-strike collision [48]. This highlights the importance of dynamics during normal gait. The stability of gait can be predominantly observed in the sagittal plane. Step length is relatively insensitive to anterior-posterior perturbations. The collision losses contribute to stability as well as the angle between the legs that increases near the end of a step. Large perturbations, however, can change the step length and can cause substantially shorter steps that dissipate less energy. Insensitivity to lateral perturbations is less pronounced, and therefore, requires active feedback control. Kuo et al. suggested that medio-lateral stabilization can be achieved by [48]:

- moving the torso from side to side,
- producing active eversion/inversion moments at the ankle, or
- changing the foot placement actively by using hip abduction and adduction.

Since lateral stability requires active control, it also requires energy. Energy expenditure can be decreased if step width and its variability is reduced too [52]. The connection between step width variability and stability is in good agreement with the results of other research studies [53, 54, 55].

Dynamic walking demonstrates that the work performed on the COM for redirection, the work required to move the legs relative to the body and lateral stabilization are the main contributors to the cost of energy of normal gait. Metabolic energy is being used to perform work in order to make a system move. Also negative work requires a considerable amount of energy. Using such a significant amount of work and conserve energy at the same time is not necessarily a contradiction. It is possible for antagonistic muscles to perform an equal amount of positive and negative work on a pendulum. This would result in zero work being performed on that pendulum while the pendulum would be still able to conserve energy through the exchange of mechanical energy.

2.2.3 Energy transfers in a multi-segment system

The dynamic walking model demonstrated that work and energy flows are important to understand the causes of movements. Dynamic walking is mainly focusing on the COM and ignores energy transfers and storage between segments. This section will look more closely into energy transfers of a multi-segment system like the human body and the role of muscles.

A body segment can only change its energy if there is a flow of energy. There can be a change from potential energy (PE) to kinetic energy (KE) or vice versa, or, regarding mechanical energy, it means that work needs to be done. This is dictated by the law of conservation of energy that applies to all points in the body and at all instances in time. The algebraic sum of all energy flows must equal the energy change of a segment. Even though work and energy have the same units, namely Joule (J), they are not the same. Energy is a measure of the ability of a body to do work and work is a measure of an energy flow from one body to another. If work is done, there is the transfer of energy moving an object by a force and the object has covered a distance in a specific direction. For example, a muscle can perform work on a segment if energy flows from the muscle to the segment. That means that the muscle provides a force to move the segment.

Joint power and work

Muscles are the main source of mechanical energy generation and absorption. The human body uses metabolic energy for the muscles to perform work and to produce force when there is no work performed. The total body energy increases when muscles do positive work. Consequently, the total body energy decreases when muscles perform negative work.
When considering the muscle power P_m in Watt, it can be calculated from the net muscle moment M_j in Nm and the joint angular velocity ω_j in rad/s.

$$P_m = M_j \omega_j \tag{2.1}$$

Therefore, concentric muscle contractions perform positive power if the muscle moment acts in the same direction as the angular velocity of the joint. If the muscle moment acts in the opposite direction to the angular velocity of the joint, negative power is performed. Eccentric contractions usually happen if an external force is acting on a segment that creates a moment that exceeds the muscle moment.
The generated energy W_m transferred from a muscle to segments is reflected by the integral of the power over the time of the contraction. This shows the net work done by the muscles.

$$W_m = \int_{t_1}^{t_2} P_m \, dt \tag{2.2}$$

The time interval needs to be chosen depending on the polarity of P_m to a calculate the total positive and negative work W_m. If done incorrectly, there could be a significant amount of work done that is not reflected in the total work calculated. When moving within a given time interval with an equal amount of positive and negative work, the net mechanical work is zero.

Calculating work from net joint moments includes any internal and external work that is done. Any external power is automatically reflected in increased joint moments. Nonetheless, there is also a shortcoming. The net moment M_j results from all antagonist and agonist muscle activity, and thus, cannot account for co-contractions. For example, if $M_j = 50Nm$ and $\omega_j = 3rad/s$, the joint power would result to $150W$. In case of a co-contraction, an antagonist might provide a resisting moment of $20Nm$. To match the $150W$ in Joint power, the agonist would need to generate $150W + 3rad/s \cdot 20Nm = 210W$ $(M_j = 70Nm)$ while the antagonist is absorbing energy at a rate of $3rad/s \cdot 20Nm = 60W$. This shows that the net power and work calculation underestimates positive and negative work done by the muscles groups at each joint. In order to overcome this problem the contribution to the net moment of each muscle would need to be known. Current measurement techniques do not allow for this task to be completely solved. For example, there are 15 major muscles responsible for the sagittal plane moments around the ankle, knee and hip joints. All three moments control the knee angle during stance phase because of the kinematic chain . This results in a indeterminacy when it comes to the relation of angle changes to moment patterns or combinations of muscle activity [2]. Another work component that is not accounted for are isometric contractions against gravity. These contractions require metabolic energy but because there is no joint movement ($\omega_j = 0rad/s$), calculated power and work components result to zero. These effects must be accounted for in patients since their muscle activation patterns are disturbed and the occurrence of disregarded co-contractions could be significantly higher than in normal gait.

Energy flow and storage

Energy exists in three forms within body segments:

$$\begin{aligned} E_s &= PE + translational\, KE + rotational\, KE \\ &= mgh + \frac{1}{2}mv^2 + \frac{1}{2}I\omega^2 \end{aligned} \tag{2.3}$$

A body or a segment can also exchange energy within itself and still maintain a constant total energy level. Body segments usually contain all three energies in different combinations any point in time during a movement. There can be energy exchanges within a segment or between adjacent segments. The pendulum conserving properties of the COM described before, can also be seen in the energy transfers of the head, arms and trunk (HAT) that has two peaks of potential energy at each stride. At these instances, the HAT shows the kinematic energy at its minimum. The body falls forward and the HAT accelerates while reducing its height, and thus, reducing its potential energy (figure 2.6).

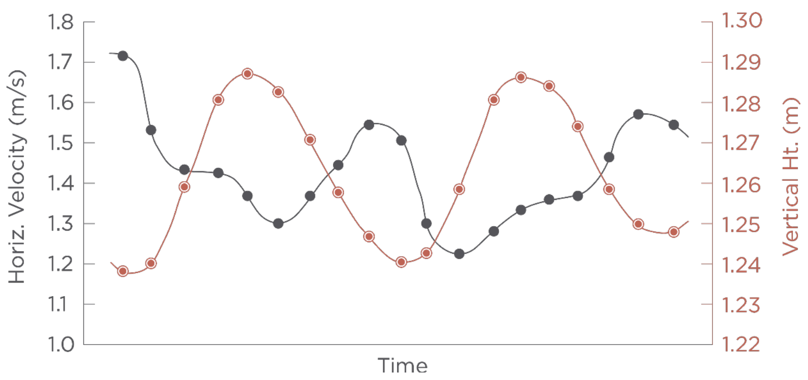


Figure 2.6: Horizontal and vertical COM velocity (data from [2]).

The potential and kinetic energy approximate sinusoidal waves that are almost out of phase. According to Winter, the transfers of energy within body segments are characterized by these opposite changes in potential and kinetic energy. To calculate the power balance within segments, all energy that enters or leaves the segment needs to be considered. This includes active transfers by muscles including absorption and generation and passive transfers across joints at the proximal and distal ends. The passive transfer of mechanical energy between to adjacent segments is possible if there is a translational movement of the joints. Work is done if there is a force displacement through the proximal and distal joint center. The process is completely passive. This is reflected in the algebraic sum of all power flows that results to zero. Consequently, the total body energy remains constant. According to Winter, this mechanism plays an important role in the conservation of energy during movements. This is due to its passive characteristics that does not require any additional muscle activity. According to that theory, this is the main reason why the swinging foot and leg lose energy during normal gait. The energy is transferred upward through the thigh to the trunk where it is conserved and converted to kinetic energy to accelerate the upper body forward [2].

The method of calculating the sum of segment energies underestimates the simultaneous energy generation and absorption at different joints, e.g. as it occurs at heel-strike (collision). The reason is that energy transfers can only be calculated within each segment and related to adjacent segments. Therefore, the resulting work can only serve as low estimate of the positive and negative work done by the human motor system.
Siegel et al. conducted a detailed analysis on the relationship between joint moments and mechanical energy during normal gait. The results suggest that energy is added to the trunk during stance phase by the plantar-flexor moment. This is even the case when the ankle power is still negative. They also found that the energy transfer becomes less effective with increasing knee flexion angles. This indicates that the flow of mechanical energy is dependent on the position of the joints and the sign and magnitude of joint moments which affects the joint stiffness at that particular time. It has also been shown that ankle and hip moments have opposite effects on the energy of the segments. It is assumed that moments with opposite energetic effects are generated to control the mechanical energy flow within the body during normal gait [56]. This might have similar functional effects comparable to co-contractions.

2.3 What makes walking efficient?

2.3.1 Mechanisms and features enabling efficient movements

The previous sections presented three theories that highlighted the main mechanisms and features of what makes normal gait an efficient locomotion task. Figure 2.7 summarizes these findings. The repetitive cycle with its time shift between the left and right leg has been described in chapter 2.1. Normal gait produces a symmetric pattern with respect to step length, step width and functional leg length. Each leg goes through the same sequence with similar kinematic parameters.

This, in turn, enables smooth trajectories of the COM and the biological joints. Constant stopping and starting would result in higher energy demand for the muscles to produce force, and consequently, in jerky movements. Another means of preventing jerky movements are presented in chapters 2.2.1 and 2.2.2. The ankle and foot complex acts like a wheel with its roll-over shape (ROS) and creates a near circular rocker. This mechanism is also known to reduce the energy dissipation at heel strike collision. Ankle push-off and powering the hip are also used to compensate the dissipation to ensure a smooth redirection of COM.

Another important characteristic is the conservation of energy as described in 2.2.2 and 2.2.3. During normal gait, this is achieved through exploitation of the pendulum dynamics and segment energy transfers. The constant change of potential and kinematic

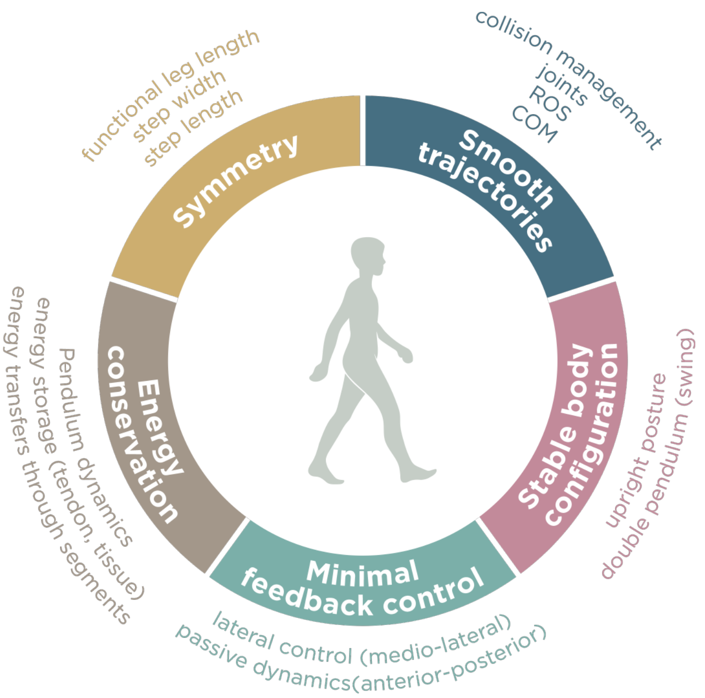


Figure 2.7: Main mechanism and features of normal gait.

energy reduces the work required during normal gait. The body is also able to store some of the energy during energy dissipating events. This energy is then released later when there is a need for power generation. This is the case at the ankle joint, for example. During early stance, the Achilles tendon stores some of the negative energy and releases it later to support the plantar-flexors at push-off.

Another characteristic to achieve efficient walking is the reliance on passive dynamics that provide a stable basis with minimal control. The is especially the case within the sagittal plane in anterior-posterior direction. Small disturbances are dissipated during each stride. The stability in the medio-lateral direction is more limited and requires active foot placement and hip work. The active feedback control needed, however, is still energy efficient especially if the step width is narrow and its variability low.

Yet another component of efficient walking is a stable body configuration. Maintaining an upright posture is crucial during stance phase. The body moves distinctively so that the ground reaction force vector only creates small moment arms towards to biological joints. This minimizes muscle force requirements for maintaining an upright posture and therefore energy expenditure. The double pendulum leg configuration during swing also reduces muscular effort. Swinging the leg when the knee is bent reduces the inertia and allows to clear the ground sufficiently.

All the mechanisms and features affect each other. If only one cannot be performed by the human body, walking will require more physical effort. For example, if there is a difference in step length, then there are also negative effects on the execution of smooth trajectories and energy conservation. As a consequence, collision management and energy transfers are not performed optimally anymore. This, in turn, affects stability of gait and will increase the energy demand related to active feedback control. It is also likely to affect the effort needed to keep a stable body configuration since the changed kinematics require increased muscle force. There is a strong interdependence between symmetry, smooth trajectories, stable body configuration, minimal feedback control and energy conservation.

2.3.2 Reduction of metabolic energy expenditure

The previous section summarized mechanisms and features that make normal gait such an efficient locomotion task without defining what that efficiency is or how it is measured. Efficiency refers to the conversion of metabolic energy to mechanical energy. The skeletal muscles use the chemical energy from the hydrolysis of adenosine triphosphate (ATP) to produce work. When a movement occurs, energy is also dissipated into heat. When the human body produces work, energy efficiency reaches approximately 30%. The remaining energy is dissipated as heat. When walking at comfortable speed, the human body is approximately 24% efficient [3].

$$efficiency_{mech} = \frac{work_{mech,int} + work_{mech,ext}}{cost_{metabolic} - cost_{metabolic,zero-work}} \tag{2.4}$$

Equation 2.4 shows that the mechanical efficiency is the internal and external mechanical work performed divided by the metabolic cost [2]. The metabolic cost needs to be subtracted by the overhead costs that are not related to the actual task. In walking, this would be the resting metabolic cost.

This definition can easily lead to wrong assumptions when it is seen as a measure of how well an individual converts biochemical energy into work. In other words, it describes the amount of useful energy that derived from a given energy supply. Winter demonstrates that in an example [2]. An unimpaired person does $100J$ of work per stride and the associated cost is $300J$ per stride. This individual then walks with an efficiency of 33%. If this is compared to an impaired person that uses $200J$ per stride at a metabolic cost of $500J$ per stride, the resulting efficiency of 40% would seem superior to the unimpaired person walking. However, the impaired individual is quite inefficient in generating an effective movement pattern, and thus, has a higher metabolic demand (metabolic consumption).

Gait or movement economy, rather than efficiency, describes this phenomenon more appropriately. Economy refers to the distance that can be traveled in relation to a specific amount of energy supplied. This is reflected in the metabolic cost of walking which is defined as energy expended per meter traveled and is measured in $J/kg/m$ or $cal/kg/m$. Metabolic cost is the physiological work (E_M) that is required to perform a task. Another measure related to energy expenditure is the physiological power (E_W)

measured in $J/kg/min$ or $cal/kg/min$. It reflects the level of physical effort. Dividing the physiological power by the speed of waking results in the physiological work. Consequently, walking can be made more economical if the physiological work, i.e. metabolic cost, is decreased by reducing the physiological power at the same speed or by faster walking speeds while spending the same power.

Figure 2.8 depicts physical power and work behavior during normal gait. The physiological power increases with faster walking speeds. The physiological work, on the other hand, is u-shaped and has an optimum at a certain speed. It has been shown that the optimal speed is adopted automatically when a person walks in a natural, self-selected manner and is about $80m/min$ [57, 58, 59]. Changes in step frequency or step length, inclinations, terrain and disabilities are known to affect energy expenditure and the related speed optimum [3].

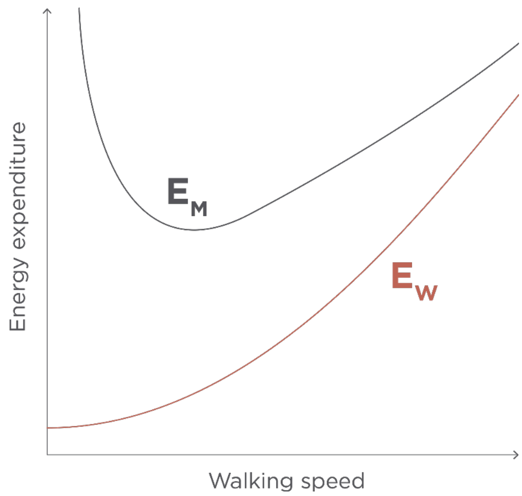


Figure 2.8: Energy expenditure per distance traveled and per time (data from [3]).

2.4 Devices to reduce metabolic energy expenditure

After identifying the mechanisms and features that make normal gait efficient and economic, one question remains: Can normal gait be more economic by using external devices that directly support the movement such as exoskeletons?

Before answering this question, it is important to point out that there are means of reducing the metabolic cost, e.g. by using a wheelchair. However, the versatility of these devices does not compare to the human legs. This is already one indicator that it is not our sole purpose to reduce energy expenditure. Task constraints always need to be considered too. If the main goal would be to cover a certain distance as quickly as possible, it would require maximum speed. This would result in an energy expenditure above the one of the optimal speed.

For the purpose of answering the question if walking economy can be improved, I only consider devices that were able to break the metabolic cost barrier. These devices are able to reduce the energy cost of walking below of what is required by the body when not using the device. Other devices that extend human capabilities and strength or increase the ergonomy of certain task are not considered [24]. They are unlikely to provide insights related to energy saving mechanisms of normal gait. In order to develop technology for patients and their predominant goal of walking efficiently again, it is imperative to understand the design features of devices that can break the metabolic cost barrier and how they influence human movement. The principles applied by these devices could potentially be used in robotic applications for patients too. One could argue, however, that there are other options of bipedal ambulation with a benefit regarding walking economy when used with an external aid. But this does not seem to be the case for two reasons. Firstly, it has been shown that the inverted pendulum walk is energetically optimal [60]. Secondly, movement and walking supporting devices should also be able to allow physiological movements to prevent secondary conditions.

Sawicki et al. identified thirteen publications that presented improved walking economy using an exoskeleton compared to not using any device [61]. Figure 2.9 compares these devices and shows the metabolic change that they have achieved.

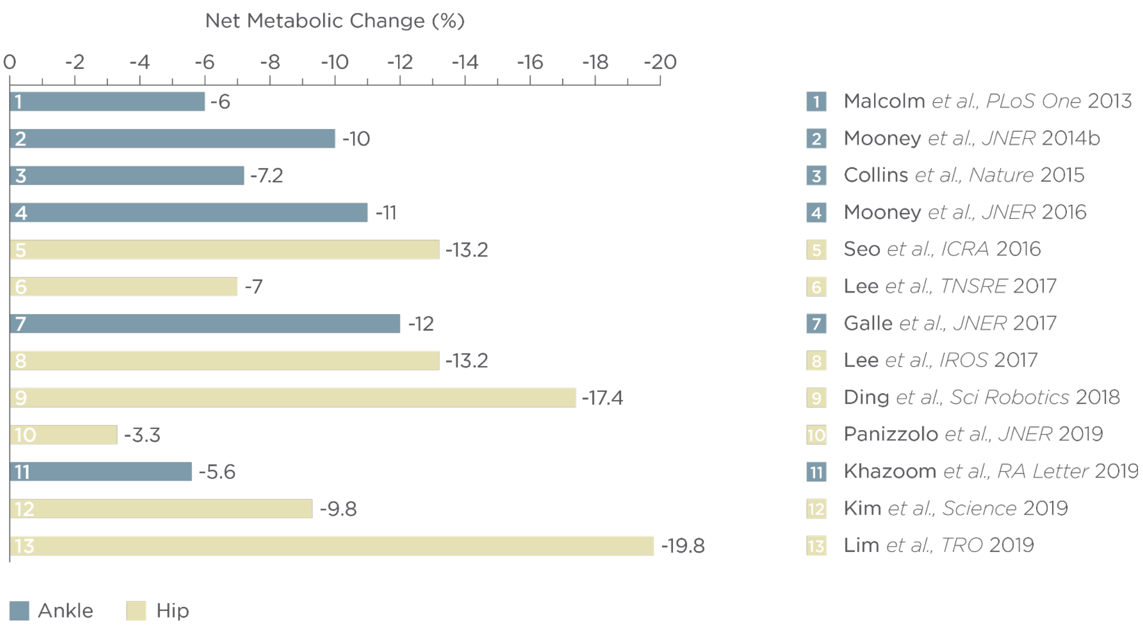


Figure 2.9: Devices that broke the metabolic cost barrier in normal gait.

These exoskeletons do only support a limited of amount of joints. This is either the hip [62, 63, 63, 64, 65, 66] or the ankle joint [67, 68, 69, 70, 71, 72, 73]. Although the exoskeltons are only attached to one joint, there is also force transmissions to other joints too. This is the case for both stance and swing phase. Hip extension, for example, can create an extension moment at the knee joint by moving the thigh segment posteriorly. When designing exoskeletons with the aim to improve movement economy, these inter-segment relations need to be considered too. Otherwise a specifically designed moment profile for one joint could generate a metabolic penalty because it is generating a restricting moment at another joint.

These devices do not enforce distinct walking patterns on the user, but rather support by providing a controlled and well-timed supply of energy. Both active [62, 63, 63, 64, 66, 67, 69, 70, 71, 72, 73] and passive [68, 65] strategies have been implemented. The maximum provided power of these devices is usually a fraction of the biological power [61]. Nonetheless, the timing and magnitude of support force seems to be crucial to affect movement economy positively [70, 74, 75]. That does not mean that a higher assistance magnitude guarantees a higher metabolic benefit [68, 71, 75]. The provided moments patterns do also not necessarily correspond to the biological moment patterns to provide a metabolic benefit [75]. The control approaches used are manifold [28] and include impedance controllers that rely on the user's kinematics as well as time based controllers that predefine certain patterns and adjust these patterns according to control parameters. The active devices use predominately electric motors because of their controllability and weight performance ratio [76].

Independent of the control strategy, using these devices may not bring an immediate benefit to the user. Novice users usually require some practice to adapt the nervous system and learn the new optimal coordination strategy [77, 78, 79]. This reorganization of coordination and the related initiation of energy optimization also depend on physiological and neurological differences. However, once the nervous system learned to predict the new optimal gait, the user can return to the new pattern quickly [80]. This might be comparable to using a bicycle. It takes some effort to learn to use this very economic way of transport and once learned it remains a life-long skill.

Exoskeletons that support the ankle joint support the push-off phase and devices that actively aid with hip movement support continuously flexor and extensor muscles, respectively only hip flexion passively. These exoskeletons usually weigh less than $6kg$ and have very low distal mass. The highest reduction in metabolic cost has been observed in an active hip exoskeleton with a reduction of 19.8% [66]. The highest reduction for an active ankle exoskeleton has been 12% [71]. All the tests have been performed around preferred walking speed which corresponds to the optimum of physiological work. The reason why hip exoskeletons have achieved greater reductions could be due to the fact that distal muscles are more efficient compared to proximal muscles [81]. Another reason could be related to the addition of distal masses that precipitate a greater metabolic penalty compared to proximal attachments [82].

A common challenge of these devices is to interface them with different body shapes in comfortable way. Multiple degrees of freedom, deforming tissue and sensitive points of pressure make it challenging to transmit forces reliably. To overcome these challenges, research teams use orthotic fabrication techniques, textiles and remote force transmission techniques.

Even though the main design features of exoskeletons with metabolic benefit have been demonstrated, it remains unclear how these devices have achieved that goal. It does not appear that they fundamentally change the mechanisms of efficient walking shown in figure 2.7, but rather provide small changes to support the muscles that are the main drivers for an efficient locomotion pattern. After all, muscles and the related force production are the main consumers of metabolic energy. The presented exoskeletons have been engineered to reduce the user's mechanical work or moments at the biological joint. But this cannot be the main reason for a reduction since this approach could also cause inefficient co-contractions or simultaneous energy absorption and generation at different joints. These biomechanical parameters cannot sufficiently describe the underlying muscle dynamics. This is also the reason why Beck et al. suggests that the alteration of the contractile dynamics are the main reason for metabolic benefit of exoskeletons; specifically by reducing the volume of muscle that is actively cycling actin-myosin cross-bridges [83].

2.5 Essential requirements derived from normal gait

Given that these devices reduce the energy demand required for normal walking, can they also achieve the same results in patients?

ID	Requirement
RQ1	Light-weight design $(< 6kg)$
RQ2	Minimal distal masses
RQ3	Adjustability to different body shapes
RQ4	Consideration of tissue deformations
RQ5	Optimization of inter-segmental force transmission
RQ6	Intuitive control to facilitate motor learning
RQ7	Avoidance of sensitive pressure points

Table 2.1: Requirements derived from exoskeletons that were able to break the metabolic cost barrier

This is unlikely since pathological gait deviates from normal walking. Even if the interface can be fitted to a patient's body sufficiently well, the controllers might not work as intended [84]. Features that are present in normal gait might not be detectable in patients. Furthermore, it is hypothesized that these devices bring a metabolic benefit by reducing muscle volume. This, however, can only be achieved if the muscle activation pattern is not changed and behaves in an expected manner. Patients will have altered activation patterns. They are also likely to struggle with the execution of the mechanisms of efficient walking shown in figure 2.7. Deviations and potential compensations are not addressed by the presented devices. It might be beneficial to provide support that is able to restore these mechanisms. Nonetheless, the essential design features of the described exoskeletons present general requirements and principles that also apply to devices that are specifically developed for patients. Table 2.1 summarizes these requirements. The devices need to be light-weight and have low distal mass to ensure that the burden because of the added mass is kept minimal. It has also been noted that the adjustabliliy to different body shapes needs to be high in order to ensure reliable force transmission. This reliability might be further affected by tissue deformations and sensitive pressure points, and thus, need to be taken into consideration. When developing new device, it is important to regard inter-segmental force transmissions which can cause undesired moments around joints that are not primarily targeted. For example, during stance phase, an external extensor moment can generate and knee extensor moment too. These interrelations need to be specifically addressed during the development process. Additionally, it has been reported that the efficient use of the presented devices requires significant time to familiarize. I conclude that this might be also the case for devices that compensate impaired movements. Therefore, a reliable and intuitive control approach might help patients to adapt to an exoskeleton.

Chapter 3

Assisting pathological walking

The previous chapter identified the main mechanisms and features of normal gait. The main mechanisms are: symmetry, smooth trajectories, stable body configuration, minimal feedback control and energy conservation. It can be assumed that if any of these mechanisms cannot be performed entirely due to a disability, then the resulting movement will take more effort [59].

This chapter starts by identifying how and why patients deviate from a normal gait pattern. The causes for pathological movements (section 3.1) indicate why patients cannot use the aforementioned mechanisms that result in an efficient and economical movement. Subsequently, section 3.2 demonstrates how currently available movements aids compensate impaired movements. The properties of these devices are used to derive further essential requirements for assistive exoskeletons. As there is a multitude of different devices and support strategies, it is not possible to conclude which joints or muscles should be supported to benefit different patients and pathologies. For that reason, section 3.3 discusses two methods of identifying support strategies related to the provision of force. This refers to the decision of which biological joints should be assisted, in what direction and during which gait periods. The first method determines net joint moment synergies in unimpaired movements. It analyses combinations of joints supported by external moments through cable actuators. Different combinations are ranked according to their ability to maximize the provided work. This method, however, has shortcomings described in section 3.3.1. Therefore, another method has been developed that is based on known deficits during impaired walking (section 3.3.2). This method provided requirements related to the provision of force and concluded the set of essential requirements presented in section 3.4.

3.1 Causes of pathological movements

Maximum energy efficiency with appropriate forward movement is produced by controlled muscle contractions. The generation of muscle force requires energy. Hence, it is beneficial to utilize energy saving mechanisms. Because of these mechanisms, many muscles can act isometrically to allow an upright posture against the influence of gravity or to transfer and store energy. Energy-expensive shortening contractions of muscles are only added at specific timings to provide power for redirection of the COM and forward motion. That is also the reason why most muscle activity occurs at the beginning of swing and stance phase.

These mechanisms are often disturbed in individuals that suffer from neurological or neuromuscular diseases. Especially out-of-phase contractions can cause abnormal movements. Causes for these inefficient movements have been identified as [2]:

- co-contractions,
- isometric contractions against gravity,
- generation of energy at one joint and absorption at another, and
- jerky movements

Co-contractions
Co-contractions also occur during normal movements at a limited extent, i.e. when it is necessary to stabilize a joint by an increased quasi-stiffness [85]. In many pathologies, this involuntary mechanism can be very inefficient e.g. when flexor muscles do unnecessary positive work to overcome the negative work of extensors. Such muscular behavior is especially common in patients with hemiplegia or spastic cerebral palsy [86, 87].

Isometric contractions against gravity
During normal dynamic movements, there is little muscle activity that can be attributed to maintaining limb segments against gravity. The momentum of the body allows for a smooth interchange of energy. Slow movements, however, as they can be seen in many pathologies, require near-isometric contractions to hold limb segments or the trunk against the force of gravity. This is commonly seen in crouch gait in spastic cerebral palsy when flexed knees require excessive quadriceps activity to prevent the body from collapsing [88].

Generation of energy at one joint and absorption at another
The inefficiency of generating energy at one joint and absorbing it at another can be seen as extension of what occurs during co-contractions. Positive work of one muscle is canceled out by the negative work of an antagonistic muscle. Similarly, positive work of muscle groups acting on one joint are canceled out by the negative work being done at others. This mechanism can also be seen in normal gait during the double support phase when there is an energy increase at push-off, and coincidentally, absorption during weight-acceptance (see chapter 2.2.2).

Jerky movements
Jerky movements are characterized by a steady succession of stops and starts. Instead of efficient energy exchanges by smooth movements, there is little conservation of energy. The frequent bursts of positive and negative work have a metabolic cost and decrease the economy of movement.

These inefficiencies can lead to accelerated and increased muscle fatigue as the muscles contract continuously. This can cause a vicious cycle since fatigue leads to a loss in force, pain, tremor and a decline in work output making it even more difficult to remain an efficient gait pattern [2]. The underlying pathological mechanisms for those inefficiencies can be grouped in five categories [34]:

- deformity,
- muscle weakness,
- sensory loss,
- impaired control, and
- pain.

Deformities
Deformities hinder the patients to attain normal posture or ranges of motion because of limited passive mobility. This can be caused by rigid or elastic contractures that represent a structural change in the joint capsule, ligaments or in the connective tissue of muscles [34].

Muscle weakness
Muscle weakness refers to insufficient muscle strength. If patients only suffer from weakness while sensation and selective neuromuscular control are intact, they can modify muscle timing to maintain an upright posture during walking. Each major muscle group has a postural substitution [34]. For example, weak quadriceps can be compensated by hyperextension during walking, or weak hip flexors can be compensated by hip abductors causing hip circumduction. This, however, presents a reduction in movement economy and likely requires more physiological work.

Sensory loss
As already presented in chapter 2, impaired sensory feedback can negatively affect stability and results in a higher step variability. The increased variability has a metabolic cost that requires more metabolic energy.

Impaired control
Spasticity is an overreaction to stretch. It impairs motor control and impedes selective motor control. This usually forces the body to adapt primitive locomotor patterns that change the phasing of muscle activity significantly. As a consequence, patients can only walk voluntarily by exploiting mass flexor and extensor patterns. These patterns prevent sufficient energy conservation and smooth trajectories which increase the energy demand significantly [34].

Pain
When a patient experiences pain during walking, it is common to find alternate muscle activation patterns and kinematics that reduce the pain. This might affect the ability to perform the mechanisms that enable efficient movements, and consequently, it is likely to increase the effort required to move.

Figure 3.1 summarizes compensation movements during walking and also connects them to the underlying pathological mechanisms that cause them.

Gait disorders mostly result in decreased stride length and cadence. The duration of the stance phase is usually also increased. These characteristics attribute to a decreased walking speed. Slow walking speeds and an wide walking base can be seen in many gait disorders. Kinematically, this results in a reduction in the total range of hip flexion and extension, knee flexion and ankle plantar-flexion during push-off [1].

3.2 Devices to support pathological movements

Compensation strategies can enable walking in patients with neurological and neuromuscular disorders. However, this comes at the cost of slower walking speeds and higher energy expenditure. Independent of the mechanisms of efficient and inefficient walking, there is a set of fundamental requirements that permit bipedal gait. Pathological gait is only possible if the locomotor system can accomplish four functions [1]:

- the body weight is supported by each leg without collapsing
- static and dynamic balance is maintained; specifically during single leg support
- the swinging leg is able to advance to a position where it can takeover the supporting role
- sufficient power is provided to move the limbs and advance the trunk

Patients often have to use compensatory movements to substitute for lost functionality. This can enable them to fulfill the minimum requirements of bipedal gait, but also prevents them from using the mechanisms that make walking an efficient task. The causes of inefficiencies are manifold as they are patient and pathology specific. This makes it a challenging task to develop therapy and mobility aids that can benefit many different individuals and pathologies.

What options do patients currently have to improve their walking capabilities?

One option is therapy and training that can induce neuroplasticity [89]. In neuroscience plasticity refers to the ability of the central nervous system to adapt to changed environmental conditions. It has been shown that extensive use of an extremity and specific muscle groups leads to an increase in cortical representation and is usually associated with an improvement in function [90]. Individual deficits and remaining functionality are the key points when deciding for the rehabilitation goals and related therapies [91]. The core element of any modern therapy is the repetitive practice of a task oriented skill in connection with continuously increasing level of difficulty. During practice, principles of learning are considered. That is also why this approach is referred to as motor learning or motor re-learning and is known to be superior to other treatment options [92]. The goal of walking rehabilitation is to relearn the mechanism of walking to make that particular task more efficient within the capabilities of a patient.

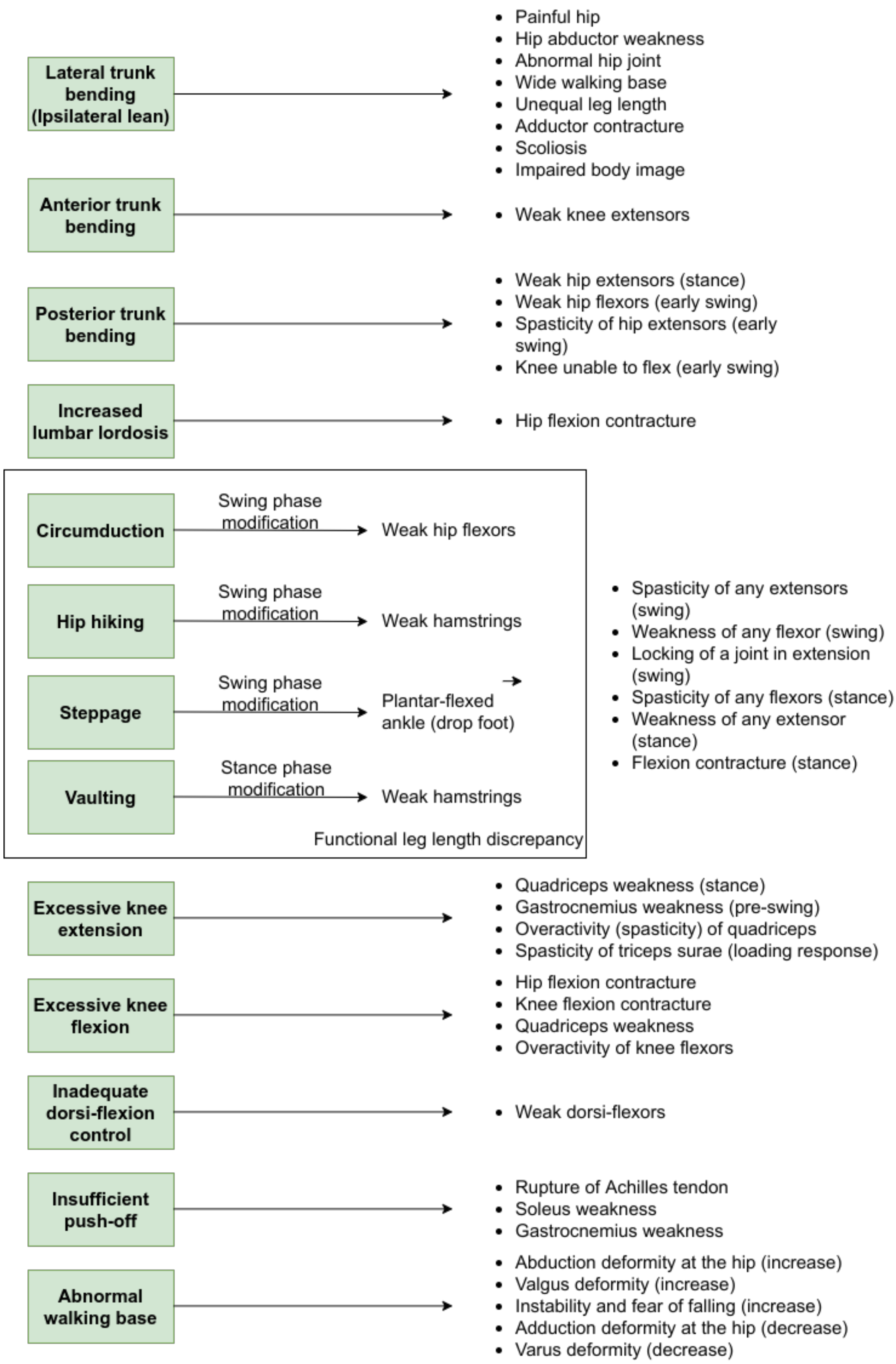


Figure 3.1: Compensation movements during walking based on data provided by [1].

The other option includes the use of mobility aids. Crutches, for example, are mobility aids that support weight-bearing and stability. They present a generic tool that can be used to help with deficits independent of the pathology. However, crutches cannot restore normal walking. They change the walking pattern fundamentally [93, 94]. It has also been shown that crutch gait is slower and has higher energy cost compared to normal gait [95]. Even though crutch gait helps with limited mobility, it cannot support regaining normal function and does not improve motor learning related to it.

Orthoses are another means of substituting lost function. Especially ankle foot orthoses (AFOs) are know to improve walking capacity measurably in terms of walking speed and energy expenditure [96, 97, 29]. They correct deficits of the ankle joint to some extent. During swing phase, an AFO prevents drop foot and helps with toe-clearance restoring some of the pendulum function in swing phase. During stance phase, they provide improved control of the tibia during the forward movement which stabilizes the knee in a limited capacity. Consequently, these passive devices also support stable upright walking. The participants in the referenced studies above are slow walkers and can be classified as household and limited community walkers [98]. Even though study participants improved their speed significantly, it was not enough to achieve a higher functional class on average. The minimal clinically important difference (MCID) of $0.1 - 0.17m/s$ [99] has not been met either. The learning of using such devices seems to be quite critical too. To use an AFO efficiently, patients require time for familiarization [29]. Devices for functional electrical stimulation (FES) that support the ankle joint in an analogous way have achieved similar results [100]. The use of other orthoses for the hip or knee joint also presented only limited success with respect to objective outcome measures [22, 101, 102, 103, 104]. Other than crutches, orthoses can provide a basis that can be used to support motor relearning of normal gait. They target certain deficits and correct deformities and support parts of the body. By doing so, these devices can improve walking functions in patients independent of the pathology.

Besides passive solutions, there are also active, robotic devices to support mobility as well as training and therapy [23]. Some devices are fully autonomous and suitable to be used as mobility aid but also as a tool in therapy [105, 106]. This is beneficial because motor relearning requires many repetitions of a task-oriented activity that exceeds what is possible in a therapy session. If a patient wants to relearn walking efficiently, then this person has to walk as much as possible. Exoskeletons are usually divided in stationary (treadmill-based) and mobile robots [23]. Alternatively, they can also be classified according to their interface. Some exoskeletons use rigid components to transmit forces to the human body. Others utilize textiles and are referred to as exosuits [107]. A classification based on technical features is flawed by its indeterminacy regarding what clearly distinguishes a devices from another. As an alternative, I suggest a classification based on the interaction with the user. This is very similar to the differentiation between an orthosis and a prosthesis. An orthosis provides support to a body part whereas a prothesis is a replacement of a body part. A similar approach can be used for function. Some exoskeletons provide support to a function like walking and other exoskeletons are able to fully substitute that function. Hence, there are exoskeletons capable of fully supporting the user's weight and moving the user's limbs entirely. These devices are able to compensate their own weight because they have a connection to the ground and the motors are strong enough to fully accelerate and decelerate the limbs. They do not rely on the skeletal structure of the user to function. I will refer to these exoskeletons as substituting exoskeletons. In contrast, there are devices that aim for partial support and require active participation. These devices can neither fully support the patient's weight nor can they move the limbs entirely. The weight of these exoskeletons is usually carried by the user and they rely on the skeletal structure to function in their full capacity. These devices will be referred to as assistive exoskeletons or exoskeletons for partial support.

Even though exoskeleton technology has made tremendous progress within the last years, current commercially available devices are not recommended by official guidelines [26, 27]. This is due to a lack of evidence that these technologies bring an objectively measurable benefit compared to conventional interventions. The issue is not related to the intended function or safety risks to patients. It seems that the cost-benefit ratio as well as clinical integration and integration into daily life are inferior to conventional therapies and the combination of crutches, orthoses and the wheelchair.

Exoskeletons that only provide partial support are likely to be advantageous for users that have residual walking capacity. They support motor relearning due to the required active participation of the user during a normal movement and because of potentially affordable prices given their lower complexity. Similar to the devices presented in chapter 2, there is an increasing number of devices for patients that are light-weight and target the support of specific functions. Figure 3.2 shows such active devices.

Only devices with available clinical data of neurological or idiopathic (i.e. elderly) patients are included in the overview. These exoskeletons support either the hip [108, 109, 110], knee [111, 112, 113] or ankle [114, 115] joint. Ankle exoskeletons are designed to assist plantar- and dorsi-flexion They support ground clearance during swing, and additionally, they are designed to restore dynamic walking, specifically the redirection of the COM, by assisting ankle push-off. They also help to control the forward movement of the tibia during early stance which provides a stable base for stance phase. Hip and knee exoskeletons can support flexion and extension; either continuously or just partially. Knee exoskeletons can support weight bearing in stance phase as well as knee flexion in late stance and early swing phase. As a consequence, they have the potential to correct stiff gait and knee hyperextension. Hip exoskeletons can restore dynamic walking during stance phase by supporting the redirection of the COM. At the same time, the also assist with knee extension when the device moves the thigh backwards. During swing, hip exoskeletons are able to support the pendulum movement of the swinging leg. All these devices can only produce limited, active support moments within the range of $4Nm$ [116] up to about $40Nm$ [117, 112]. The weight of these exoskeletons is between $2.4kg$ [116] and $6.5kg$ [110]. Whereas distal masses are kept low [118, 114].

It has been shown that hip and ankle actuation can potentially reduce metabolic cost and increase walking speed [118, 109, 110, 117]. Nonetheless, the related studies do not compare the results to a control group. The lack of a control group prevents an evaluation between the intervention of the devices and training effects. Similar to orthoses,

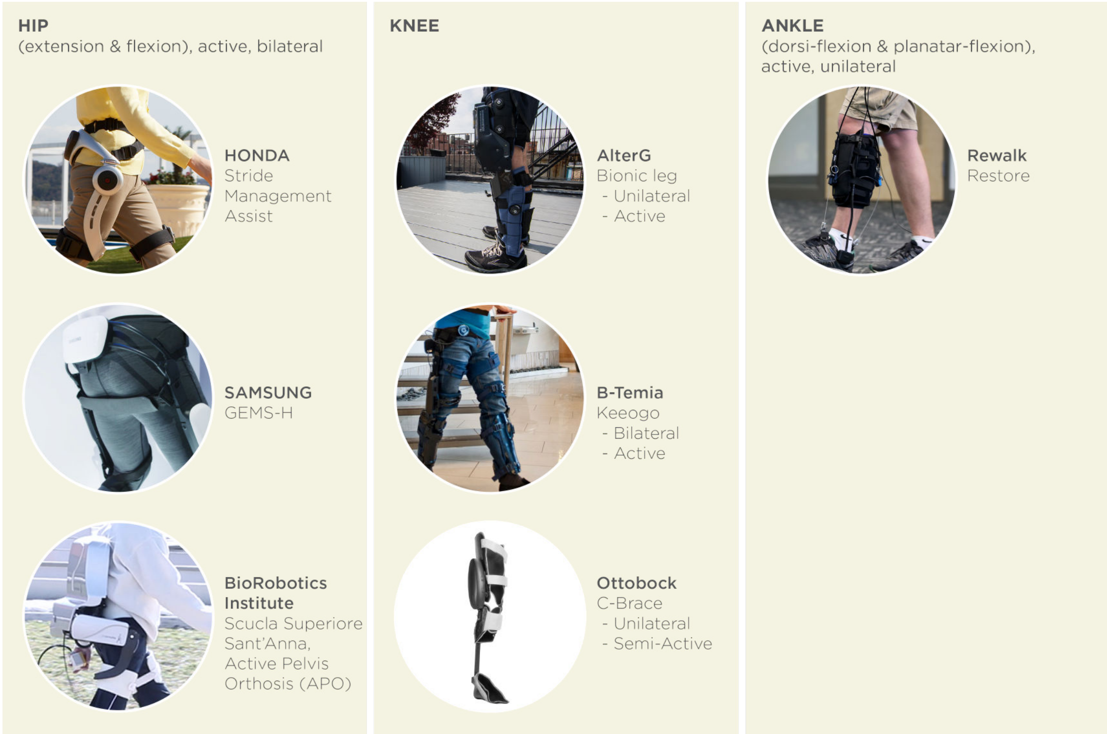


Figure 3.2: Active devices that have clinical data and compensate pathological walking partially.

these devices require a familiarization over multiple training sessions to enable patients to use them at their full capacity [119, 116, 120]. It has also been shown that speed improvements can be above the aforementioned MCID. An increase in walking speed is a common measure in clinical evaluation procedures to assess the impact of an intervention [98]. However, patients do not seem to progress from one ambulation class (e.g. household walkers) to a higher one (e.g. limited community walker). Consequently, there is no clinical meaningful improvement in function or a patient's customary level of walking ability at home and in the community. Especially ankle and hip exoskeletons seem to target higher functioning community walkers with a walking speed of more than $0.8m/s$ [117, 116]. Nonetheless, hip and ankle exoskeletons can show immediate positive effects related to kinematics, metabolics and walking speed [118, 115, 121].

Studies that compare interventions with a knee or hip exoskeleton and conventional training have shown mixed results. Tanaka et al. have shown that the Honda Stride Mangement Assist can increase walking speed significantly in a post-stroke population. However, the study included mainly fast walkers (community walkers) and the speed increase was below the MCID. Other studies with the same device could not find significant differences between the groups with respect to the primary outcome measures [120, 108].

Stein et al. conducted a randomized controlled pilot study with a robotic knee brace in individuals with chronic hemiparesis after stroke. There has been no significant differences between the groups that used the exoskeleton and the group with the exercise program [111]. A potential explanation is related to the higher distal masses of knee exoskeletons. While being more proximal than devices attached to the ankle, they usually are more complex with regard to their body attachment and force guidance across the joint, and thus, they tend to be heavier. Because these devices cannot compensate for their own weight during swing phase, they might hinder the person in taking a step. Even if a patient is supported during stance phase, the weight of the device might reduce step length, and therefore, walking speed and distance traveled in a given time. Mcleod et al. found in their investigation of a knee exoskeleton that people with a certain level of deficit respond well to the intervention [112]. Fast walkers could not gain a benefit from the device because they are assumed to have sufficient muscular strength and no problem with weight bearing. Whereas slow walkers are presumed to struggle with the mass of the device. Consequently, the patients that struggle with weight bearing but still have sufficient strength to accelerate and decelerate the mass of the exoskeleton responded with improved walking speed and endurance.

This finding is in good agreement with the relation of knee extension score and predicted walking speed [98]. Knee flexion and extension strength related to upright motor control is a significant indicator of impairment and differentiates household from community walkers. In conclusion, slow walkers that can be classified as limited community or household walkers appear to have limited capabilities to fulfill some of fundamental requirement for bipedal locomotion, i.e. supporting the body weight without collapsing and maintaining static and dynamic balance especially during single leg support. Knee exoskeletons can potentially assist these functions and enable higher speeds that improve a patient's ability to walk in household and community settings.

If the fundamental requirements of walking do not pose a challenge to the patient, then devices for the hip and ankle that are able to improve the mechanisms of efficient walking (see chapter 2) and might be suited optimally as they can help to restore symmetry, smooth trajectories, a stable body configuration, minimal feedback control or energy conservation.

3.3 Compensation and support strategies

The previous sections showed not only how and why patients are not able to perform the mechanisms and features that make normal gait an efficient and economical movements, but also presented current movement aids and their properties related to compensating impaired movements. Unfortunately, it is not possible to derive clear requirements related to the support strategy. One important question remains. If a new device for partial support is to be developed, how can a developer find the right way of assisting pathological movements? This section presents two approaches to determine a support strategy related to which biological joints should be assisted, in what direction and during which gait periods.

3.3.1 Net joint moment synergies

To exploit synergistic effects at a joint moment level, Bartenbach et al. suggest that cable driven systems can be designed while considering the moment and power behavior during movements [4]. The approach assesses all potential assistance options that a cable drive provides. 27 combinations are theoretically possible when considering the hip, knee and ankle joint with respect to extension, flexion or no support.
The approach relies on the following assumptions and prerequisites:

- The work performed and peak moment are the main indicators with respect to the choice of supported joints
- Direction changes in net muscle moments indicate the limits of synergistic moment intervals
- Polarity changes of power within the synergistic moment intervals indicate synergistic power intervals where moments and joint velocities do not change their direction
- Synergistic power intervals can only be supported if the coupled joints are either both energy generating or absorbing at the same time

Figure 3.3 shows an example of how the method works for a potential combination of the hip and knee joint during normal walking. At first, all zero crossings for the hip and knee moment were identified (1-8). This procedure then determines intervals where moment profiles are constantly flexor or extensor moments for both joints without a change. Within each interval, the joint powers are evaluated. Because of the prerequisite that a cable actuator can only extend or contract at a given time, the joint powers of two joints must both be either generative or absorbing simultaneously. The resulting interval 2, for example, must further be divided in interval $2a$ and $2b$. According to this theory, interval $2a$ is not applicable because of the difference in polarity. Interval $2b$, however, would be suitable to be assisted by a cable actuator. In this particular case, the cable actuator should provide a knee and hip extensor moment during power generation of both joints from 16% to 28% of the gait cycle.

For all the intervals fulfilling the power requirement, the work performed is calculated. All combinations are then compared according to the work that is performed. This eventually results in ranking of all combinations. Finally, the given intervals are also screened for peak moments. Bartenbach et al. theorized that peak moments are crucial for the cable actuator to provide substantial assistance during particularly demanding movements. Following this method for data of unimpaired sit-to-stand transitions, stair ascent and normal gait, supporting hip and knee extension might potentially be most beneficial for a user. Although this method might give a rough estimate of joint couplings for cable actuators, the underlying theory is incomplete and has limited potential to predict optimal support for pathological movements. It can only serve as a lower boundary as it underestimates the attainable intervals to provide assistance.

The presented method is based on the directionality of net muscle moments to identify intervals that could be supported by using a single cable actuator. The net muscle

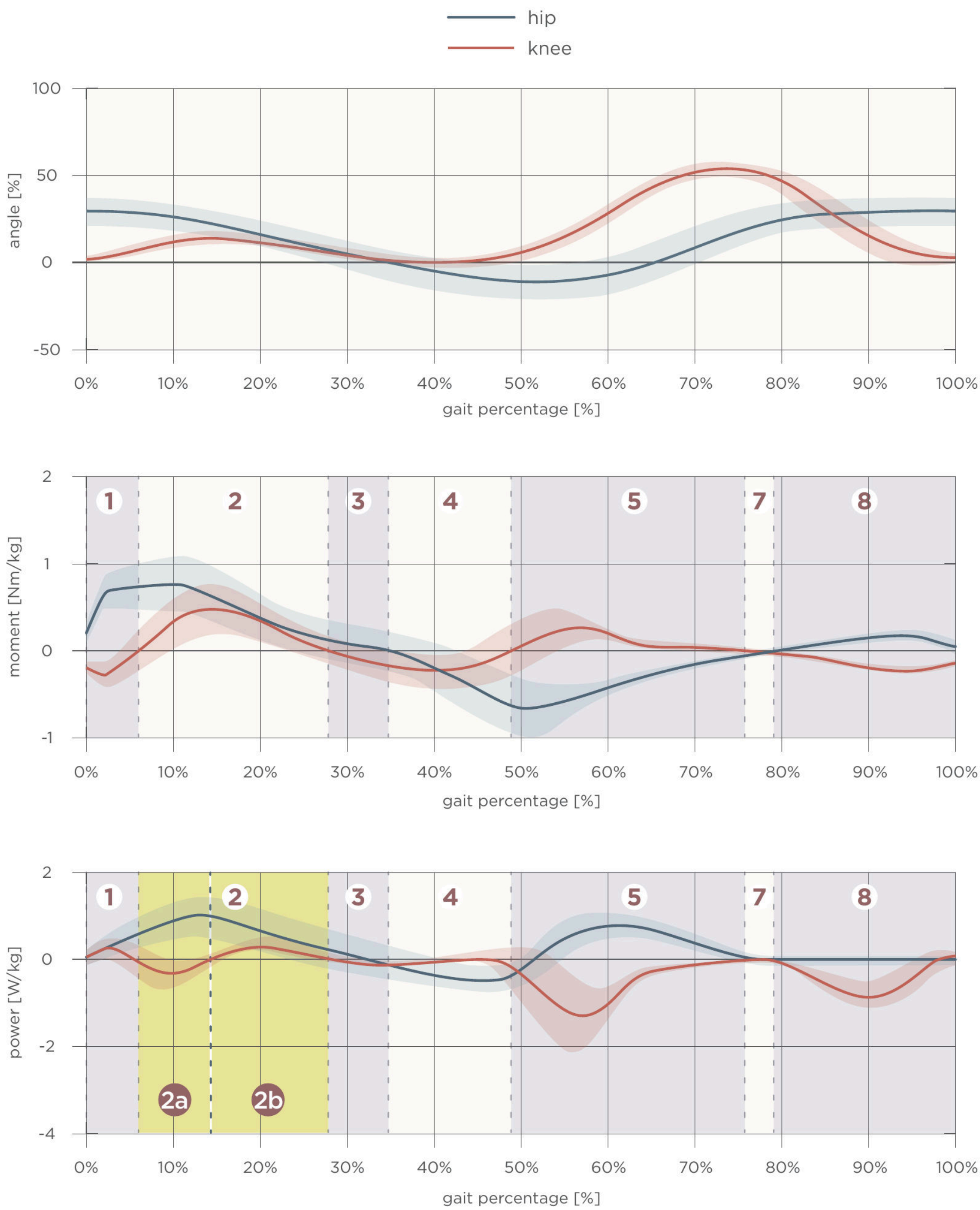


Figure 3.3: Angle, moment and power of the knee and hip joint during normal gait. Intervals have been established according to the method described by [4].

moment around the joints represents the sum of all moments created by the muscles. Adding an external actuator that can deliver about 30% and 50% of typical joint moments of an unimpaired person [4] will add to that sum of moments. Therefore, the device works as an external muscle that is not able to fully drive the moment behavior and related movement generation at a joint. If the device could provide sufficient force to fully reproduce movement without changing the dynamics of the whole system, it might be adequate to reproduce the moment curves in the identified intervals as suggested. However, this is not the case for partial support. This leads to the question of how an external moment can alleviate movement?

Movement can only be alleviated if the involved muscles produce less force and work. This is reflected in the muscle dynamics [83] and not in the net moment sum around a joint. After all, it is the muscles that require energy to contract eccentrically, concentrically or isometrically. Chapter 2.2.3 showed in detail that the net muscle moments cannot account for co-contractions. It only provides limited meaningfulness regarding muscle dynamics. Martini et al. have even shown that an active pelvis orthosis can reduce energy expenditure when tailoring the support moment according to muscle activation. This strategy resulted in an external flexion moment provided by the device during mid-swing, although the net moment around the biological joint would be an extension moment [110]. This design has been proven to be beneficial for the elderly and is in direct contradiction with the presented theory.

In the example depicted in figure 3.3, interval 1 ends with the change of the knee moment from flexion to extension. During this phase, the limb needs to support the body weight while also damping the impact caused by heel strike. The knee continuously flexes while the hamstrings provide knee flexor and hip extensors moments, the gluteus maximus contributes a hip extensor moment and the quadriceps a knee extensor moment. Antagonistic muscles stabilize the knee. Even though there is a flexor net muscle moment present, it would be still beneficial to provide an external extension moment to reduce the force requirements on the quadriceps, and thus, reduce the effort to stabilize the knee. Consequently, there is no need to have interval 1 and 2 separated in this specific example.

The theory presented by Bartenbach et al. explained further that the polarity of the joint power cannot change in a given interval. The reason for that is the inability of an actuator to contract and to elongate at the same time when spanned over multiple joints. This assumption is incorrect when it comes to compliant interfaces. Figure 3.4 shows a simple mechanical model of a segment connected to a joint. The tissue around the segment is modeled as a spring at the attachment point of the linear actuator. Muscles generate a moment M_j around the joint. At the same time, the cable actuator provides a force that is contributing to the net moment. If the connection between the cable actuator and the segment would be entirely stiff with $k_s = \infty$, then power generation of the actuator (F, v of equal polarity) would also result in power generation around the joint. This behavior automatically implies that $F_x l_s > M_j$. If the interface is not stiff ($k_s \ll \infty$) as it is expected in any human exoskeleton interface, there can be power generation at the actuator while there is energy absorption at the biological joint. The main reason is the decoupling of the joint movement and the cable actuator with respect to tissue compliance. Even though the example in figure 3.4 shows only one joint, the principle also applies to multiple joints. Hence, depending on the interface stiffness, it is possible to support joints with different polarity of power.

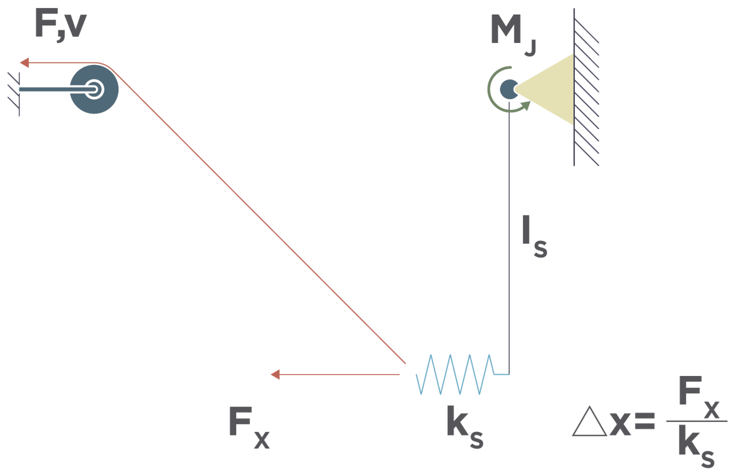


Figure 3.4: Mechanical model of a single segment connected to a joint and to a cable actuator elastically.

Considering again the example in figure 3.3. The requirement related to joint powers divided interval 2 into two distinct intervals. This separation does not seem necessary given small knee flexion angles. We have shown that high actuation forces of about $350N$ can be applied without generating a movement resulting only in tissue compression [122]. This shows that there is a disconnection between the external system and the human body because of compliance. There is also a limitation to the compliance and the disconnection as the tissue and interface compresses at higher forces. Nonetheless, within certain boundaries, it is possible for an actuator to generate energy and for a joint to absorb energy at the same time. Bartenbach et al. also assumed that the coupling of two joints restricts free movements of these joints. This assumption is not entirely valid for compliant interfaces.

The main weakness of the presented approach is that it treats the human and exoskeleton system as if the exoskeleton could drive the whole system dynamics. Their method of identifying synergistic net moments and work rates does not consider the underlying muscle dynamics and ignores the compliant interface. Chapter 2.2.3 already showed how the net moments and related work rates cannot account for co-contractions and isometric contractions against gravity. These muscle contributions also require a significant amount of energy. The claim that this method identifies peak torques in a given interval is also controversial in regards to its effect on the human body. Collins et al. have demonstrated that maximizing the provided moment can result in a metabolic penalty [74]. If an actuator can deliver about 30% and 50% of typical joint moments, it is questionable if the peak moments should be prioritized to maximize work. Beyond that, Beck et al. have argued that work provided is not the main driver behind the mechanism of reducing effort during normal gait [83]. The identified combinations are ambiguous related to their recommendation of support in the identified intervals. When the method recommends to support the hip and knee moment, it means that a single cable should be guided along the two joints that moments are generated. It completely ignores force transmission by segments within the musculoskeletal system. For example, if the method recommends to assist hip and knee extensors, it does not mean that a single cable connector needs to connect these joints. The hip extensors move the thigh postoriorly during stance. This creates an extensor moment across the knee joint. Consequently, by supporting the thigh only, there are supportive moments generated in the knee too. The proposed synergistic moment intervals indicate by no means that this combination can only achieved by a cable actuator spanned over these joints.

Bartenbach et al. concluded that a cable actuator should simultaneously support hip and knee extensor moments during normal gait, stair ascent and sit-to-stand. This result, however, is skewed because it compares a zero-work movement like normal gait to two movements that require a significant increase in body energy. According to the suggested method, the combination would also only provide very little assistance during normal gait. Two of the three movements considered require the body to move against gravity while also increasing its potential energy significantly. Therefore, high extensor activity is required to provide the required positive work. Stair ascent and sit-to-stand transitions can only exploit energy conserving mechanisms to a limited extent. They do not rely on isometric contractions as heavily as it is the case in level walking. Hence, the best combination might be movement specific and cannot be calculated as the average over different locomotion tasks.

Zhang et al. have shown that the metabolic benefit from an external device can be maximized when the assistance is individualized for a particular person [75]. This is the case despite the high similarity of normal gait between individuals. Given that small differences have a big impact on optimal assistance profiles, how can data of unimpaired individuals then serve as a basis for pathological movements? Pathological movements are pathological because they deviate strongly from normal movements. This is the biggest shortcoming of the suggested approach. For exoskeletons that take over movement, it might be beneficial to adapt normal movement profiles. However, this is not possible when providing partial support. The moments provided will add the net joint moments but they are not able to drive the body kinematics. It is to be expected that the muscle activation patterns in pathological movements are disturbed which could further hinder patients as they have limited capabilities to adapt instantaneously. Providing support according to normal activation patterns might disturb compensatory movements and adapted muscle activation patterns. There is no indication that would support the conclusion that a provision of partial and physiological moment profiles can restore normal function.

This method has significant shortcomings in terms of ability to recommend support strategies for patients. Nonetheless, it shows the difference between net-zero-work tasks like normal gait and movements against gravity that require a significant increase in body energy. Extensor moments at the hip and knee seems to play a crucial role during these movements. This finding might be beneficial when designing devices that also support other movements than normal gait.

3.3.2 Deficit based support

A patient's deficits need to be taken into account when developing suitable support strategies for pathological gait. Patients usually develop compensation strategies that enable bipedal locomotion at higher energy demands. The compensations can be seen as deviations from kinematic pattern observed in normal gait. The burden of a disease could be alleviated by addressing the pathological mechanisms for inefficiencies mentioned in chapter 3.1. Out of the five mentioned mechanisms, an external device can potentially help with three of them. Deformities could be corrected, weak muscles can be supported and impaired motor control can be improved by supporting antagonistic muscles. An exoskeleton that can only provide partial support must address the root causes to restore normal gait and support motor relearning. When considering the mechanisms of efficient movement (chapter 2.3), deformities, weak muscles and impaired motor control disturb them. Devices that can help to make a movement more efficient and economical again need to directly target the cause of the inefficiency.

Whittle [1] and Perry [34] provided an in-depth overview of common compensation strategies. I have used that information to show the most common deviations in the sagittal plane at a joint level for the hip, knee and ankle. Afterwards, I identified the affected gait phases and underlying causes. A deviation at only one joint can also cause movement changes and substitutions that affect the whole body. These common changes and functional substitutions are considered too and associated with stance or swing phase for a given deficit. Given the deviations and related deficits, support methods have been assigned according to the following set of rules:

- Weak muscles can be directly supported by an external moment around the same joint that a muscle acts upon
- Weak muscles can be indirectly supported by an external moment around other joints that the muscle does not span; but only if it is unlikely to facilitate a compensatory movement
- Deformities (i.e. contractures) can only be supported indirectly by providing an external moment to assist affected muscle groups but not the deformity itself
- Impaired motor control (i.e. spasticity) can be aided by supporting antagonist muscles in parallel with an external moment

Given all potential deficits, their causes and the derived external support methods, a simple grading scheme as been implemented. The grading scheme assigns points to potential support solutions to counteract deformities, weakness or impaired motor control according to the rules presented above. It then considers the gait periods that could be supported. Each supported gait period results in one point in the grading scheme. All potential solutions are eventually evaluated to find the one that potentially can address most deviations and deficits.

Ankle deviations

Figure 3.5 shows the main two ankle deviations: excessive plantar-flexion and excessive dorsi-flexion which also includes the lack of normal plantar-flexion. Excessive plantar-flexion reduces heel contact and causes difficulty to maintain an upright posture during stance phase. Loading response, mid-stance and terminal stance are affected by the deviation.

A potential body adaptation is the knee recurvatum. The plantar-flexion angle at the ankle joint forces the tibia posteriorly. This causes a knee extensor moment that can result in knee hyperextension. Subtalar eversion is another compensation mechanism that helps to increase the contact between the foot and the ground. Excessive plantar-flexion can be caused voluntarily to ensure a stable knee configuration when the quadriceps are weak. Other causes directly originate at the ankle joint, i.e. plantar-flexion contractures, spastic triceps surae or a weak tibealis anterior. Whereas it is challenging to correct the contracture, the spastic muscles and the weak tibealis anterior can be supported by an external dorsi-flexor moment. The spastic triceps surae requires support in mid and terminal stance while the muscle is stretched when the tibia moves forward. Since dorsi-flexor support can assist two periods, it obtains two points in the grading scheme. Support of the weak tibealis anterior is needed to prevent the foot from slapping on the ground during loading response. This results in one additional point assigned to the same support method (dorsi-flexor support, three points). When the quadriceps are weak and excessive plantar-flexion is used as a protective measure to prevent a collapse of the leg, it is beneficial to support knee extension in all three phases which results in three points according to the grading scheme.

Excessive plantar-flexion increases the effort of swinging the leg forward because of the reduced ability to clear the ground. The affected periods are mid- and terminal-swing. Substitutions and changes aim to prevent toe-drag by circumduction, hip hiking, steppage, vaulting and posterior trunk bending. The potential causes for this deviation in swing phase originate directly at the ankle and are the same as for stance phase. Dorsifexion support can help with spastic triceps surae and a weak tibealis anterior (each two points). If a contracture exists, it is more suitable to support knee or hip flexors while allowing the person to clear the ground (two points).

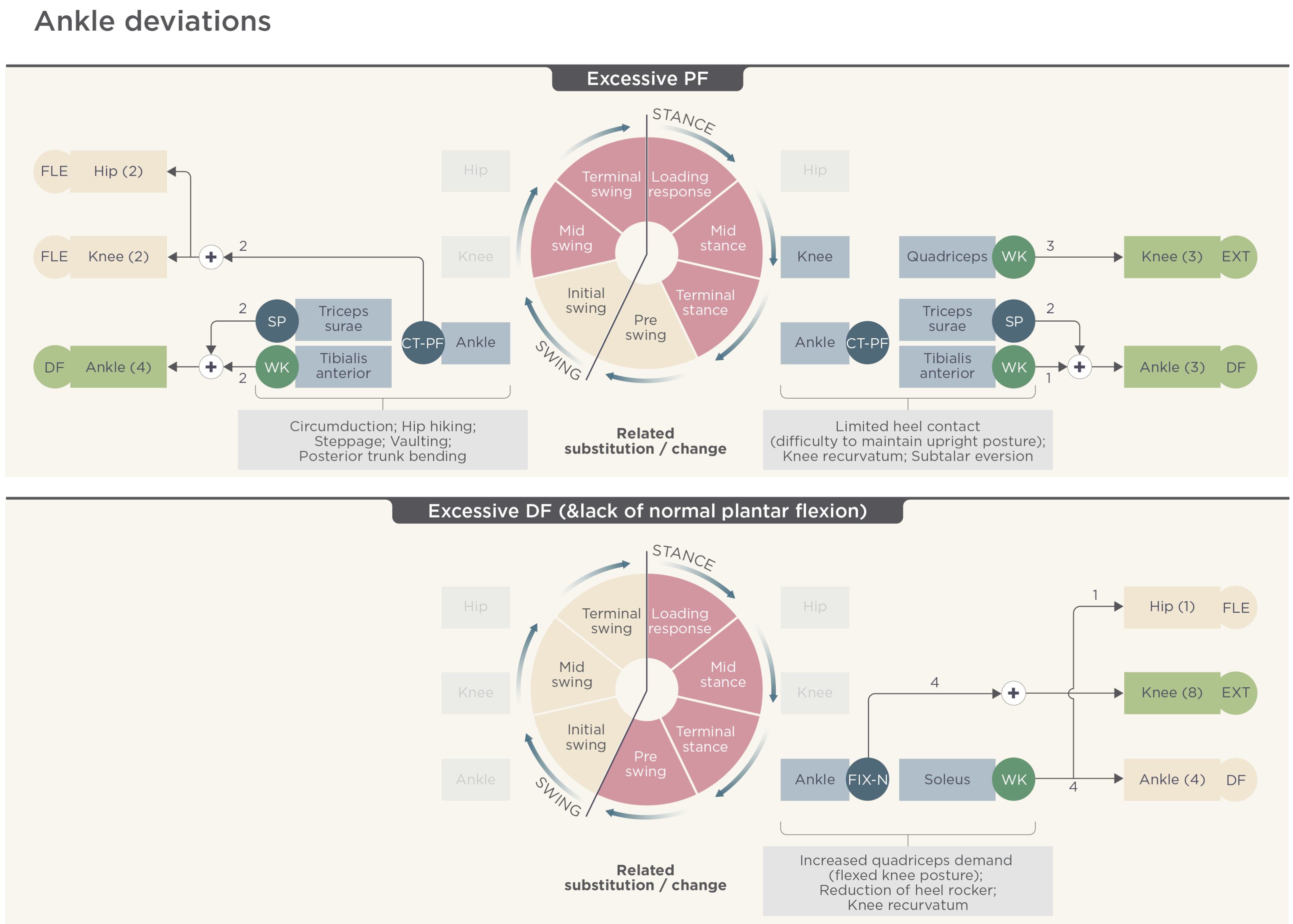


Figure 3.5: Ankle joint deviations (excessive plantar-flexion, excessive dorsi-flexion) from normal gait, related deficits and potential support strategies.

Excessive dorsi-flexion or the lack of normal plantar-flexion affects the entire stance phase. This deviation is usually compensated by an increased quadriceps demand in a flexed knee posture, reduced heel rocker or knee recurvatum. The causes for this particular deviation are related to an ankle joint fixed in neutral position, either surgically or by an orhtosis, or a weak soleus muscle. This causes an increased demand in knee extension and can be supported by an external knee extensor moment for the entire stance phase resulting in four points each for the fixed ankle joint and the weakness. The soleus can be assisted by an external plantar-flexor moment during the entire stance phase resulting in four points. Hip flexion could mainly help the preparation of the swing phase in pre-swing, and thus, obtains one point.

Knee deviations

Figures 3.6 and 3.7 depict all possible knee deviations. Inadequate flexion affects loading response and pre-swing due the lack of knee flexion. Common functional substitutions and changes include hyperextension, premature ankle plantar-flexion and anterior trunk bending to shift the ground reaction vector in front of the knee. The deviation can be caused voluntarily to compensate for weak hip extensors or by spastic quadriceps.

It can be also induced by a plantar-flexion contracture of the ankle or alternatively by weak or spastic triceps surae. An external knee extension moment can stabilize the knee and allow knee flexion without collapsing the knee joint. This is beneficial when the quadriceps or triceps surae are weak as well as when there is a ankle plantar-flexor contracture (eight points). Plantar-flexor support can help with weak triceps surae specifically (two points). If the deviation is caused by spastic muscles, the antagonists can be supported in order to facilitate knee flexion (ankle dorsi-flexion and knee flexion both two points).

Inadequate flexion affects initial swing and mid-swing negatively. This often results in circumduction, hip hiking, posterior trunk bending and ipsilateral lean to achieve sufficient ground clearance. Spastic quadriceps can prevent to knee from attaining a flexed position. This can also be causes by weak hip flexors that usually support knee flexion by moving the thigh forward during normal gait. Weak hip flexors can be either directly supported by an external hip flexor moment or indirectly by a knee flexor moment (each two points). Knee flexion creates limited hip flexion by moving the shank backwards causing the thigh to move anteriorly. Knee flexion caused by spastic quadriceps can be facilitated by supporting the antagonists externally (two additional points for knee flexion).

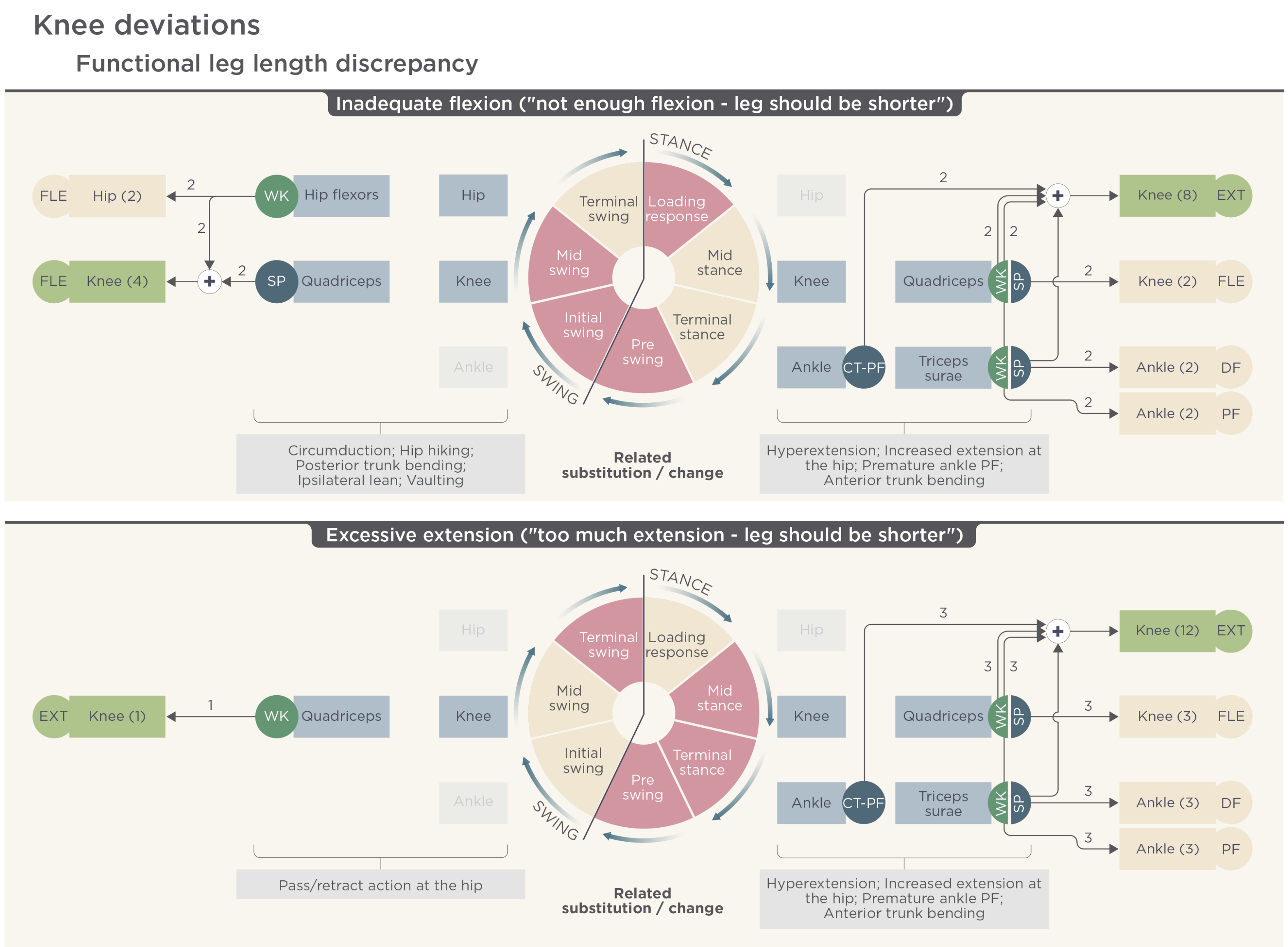


Figure 3.6: Knee joint deviations (inadequate flexion, excessive extension) from normal gait, related deficits and potential support strategies.

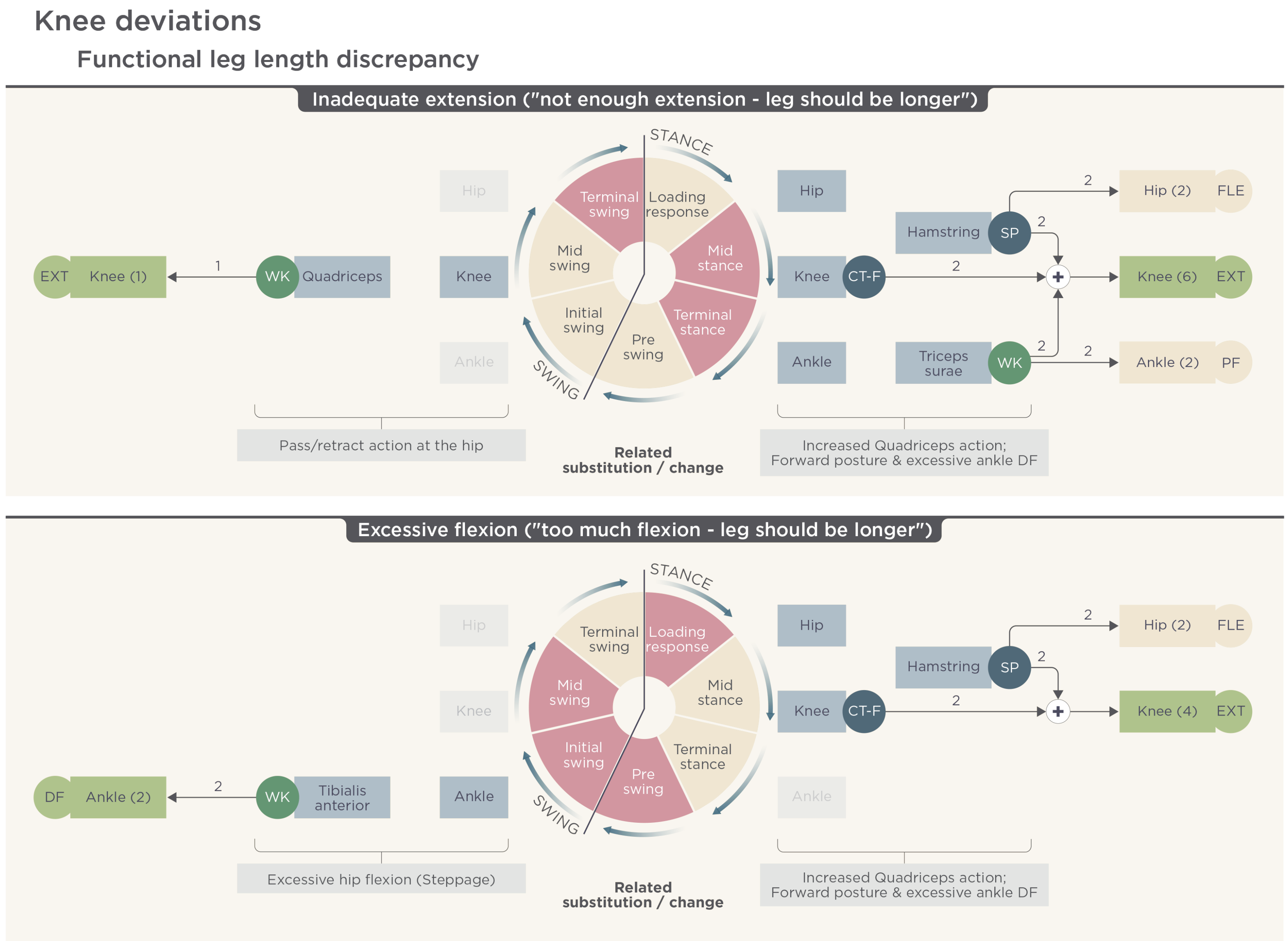


Figure 3.7: Knee joint deviations (inadequate extension, excessive flexion) from normal gait, related deficits and potential support strategies.

The same causes that lead to a reduced functional leg length induce another deviation (excessive extension) in periods where the knee is required to be extended. This particularly affects mid-stance, terminal stance and pre-swing. The corresponding substitutions and deficits are the same ones compared to inadequate flexion during stance phase. The points in the grading scheme are similar to inadequate flexion with the only difference that three periods of the gait cycle are affected resulting in higher number of points.

Excessive extension affects only terminal swing. The required functional substitution is a pass-retract action of the thigh. This causes the shank to be propelled forward. The deficit that requires the functional change is related to weak quadriceps that are not able to move the shank forward. This can be aided with an external knee extensor moment (one point).

Inadequate extension refers to instances where the knee is flexed although normal function would require knee extension. During stance phase this affects mid-stance and terminal stance. The related substitutions and changes include increased quadriceps activity, forward posture to compensate for the imposed thigh angle and excessive dorsiflexion angle to compensate for the imposed angle of the shank. The changes can be induced by spastic hamstrings, by a knee contracture or by weak triceps surae. An external hip flexion moment and knee extension moment can counteract the spastic muscles and facilitate extension (each two points). The increased knee extensor activity induced by the flexion contracture can also be supported by an external knee extensor moment (two points). Weak triceps surae are not able to control the forward movement of the shank sufficiently. This function can be substituted by either a knee extensor moment (two points) or a plantar-flexor moment (two points).

Inadequate extension affects only terminal swing during swing phase. A functional substitution for weak quadriceps is a pass-retract movement of the hip that moves the shank forward. If the movement can be done only insufficiently, the functional leg length is too short. An external extensor moment at the knee joint can support the weak quadriceps (one point).

Excessive flexion describes the opposite of inadequate extension. It also refers to a functional leg length that is too short because the knee is flexed excessively compared to flexion occurring during normal gait. Loading response and pre-swing are affected in stance phase. The functional substitutions and changes are the same ones as for inadequate extension. Spastic hamstrings can cause the deviation as well as a knee flexion contracture. Knee extension and hip flexion can support the antagonistic muscles to counteract the spastic hamstrings (each two points). An external knee extensor moment can also help with the increased load on the quadriceps caused by the knee flexion contracture (two points).

Excessive knee flexion affects initial swing and mid-swing. The related functional change is steppage. The knee flexes extensively to ensure ground clearance. A weak tibealis anterior causes drop foot. The weak muscles can be directly supported by an external dorsi-flexor moment (two points).

Hip deviations

Figures 3.8 and 3.9 show common hip movement deviations from normal gait. Inadequate extension affects mid-stance and terminal stance. Functional substitutions and changes include posterior trunk bending to move the ground reaction vector posterior to the hip joint and increased lumbar lordosis to ensure an upright posture. Inadequate extension can be caused by a hip flexion contracture, extensor muscle weakness or spastic flexor muscles. An external hip extensor moment can support the weak extensor muscles and counteract the spastic flexor muscles (four points).

Excessive flexion affects pre-swing and causes the same functional substitutions and changes as inadequate extension. Because the rules of the grading scheme do not allow for direct support of a contracture, there is no direct way of compensating this deviation. During swing, excessive hip flexion affects mid-swing and initial swing. If the deviation is caused by spastic triceps surae or a weak tibealis anterior, a common compensation is steppage. An external ankle dorsi-flexor moment can alleviate both deficits (each two points).

Inadequate hip flexion affects the entire swing phase. Common functional substitutions and changes include increased knee flexion, posterior bending, contralateral lean, hip circumduction and hip hiking. The underlying cause are weak hip flexors. An external hip flexor moment can compensate for the weakness (three points). Also knee flexors can compensate for the weakness as a flexed knee moves the thigh forward. This support is only applicable in initial swing and and mid-swing (two points).

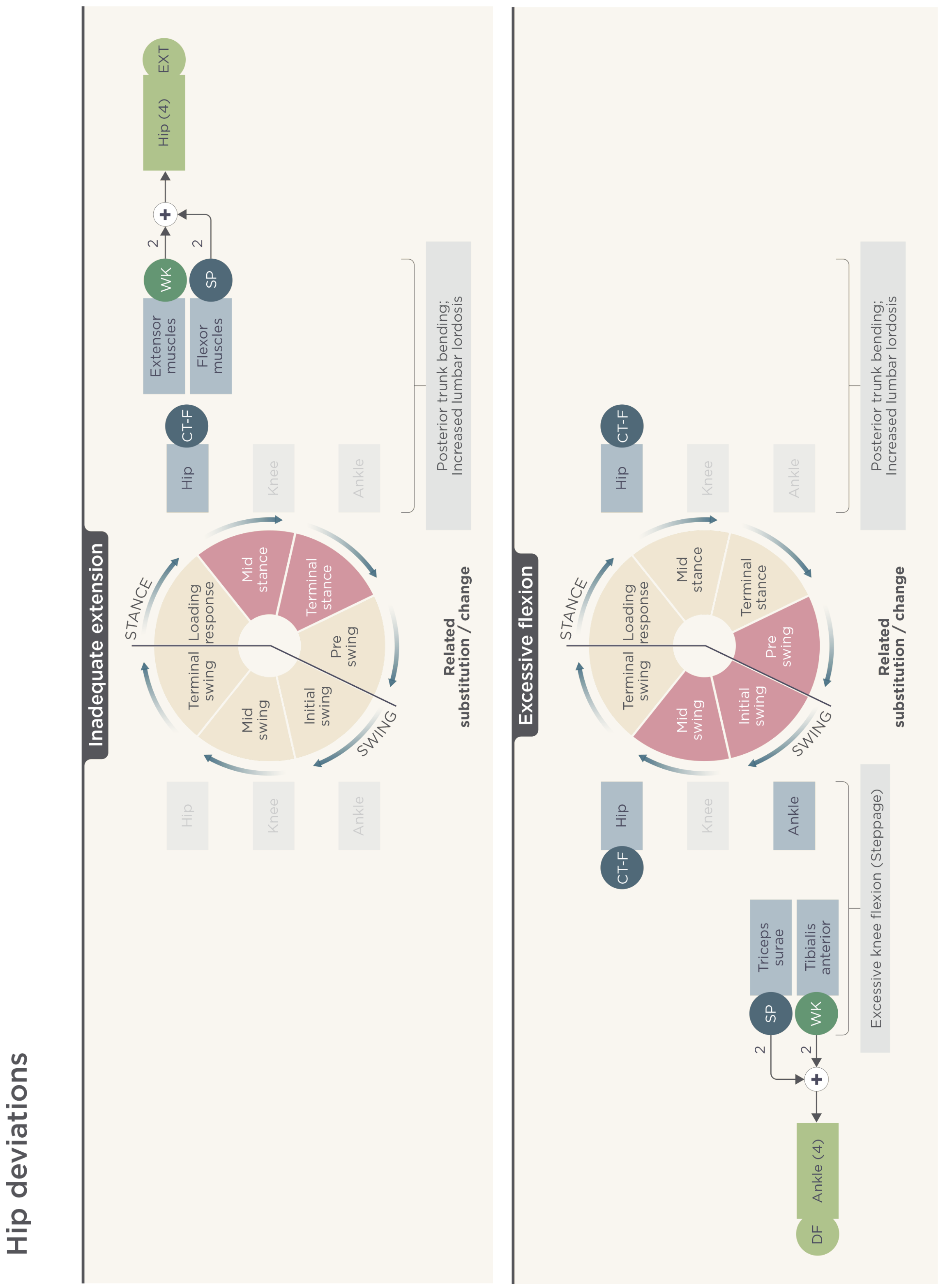


Figure 3.8: Hip joint deviations (inadequate extension, excessive flexion) from normal gait, related deficits and potential support strategies

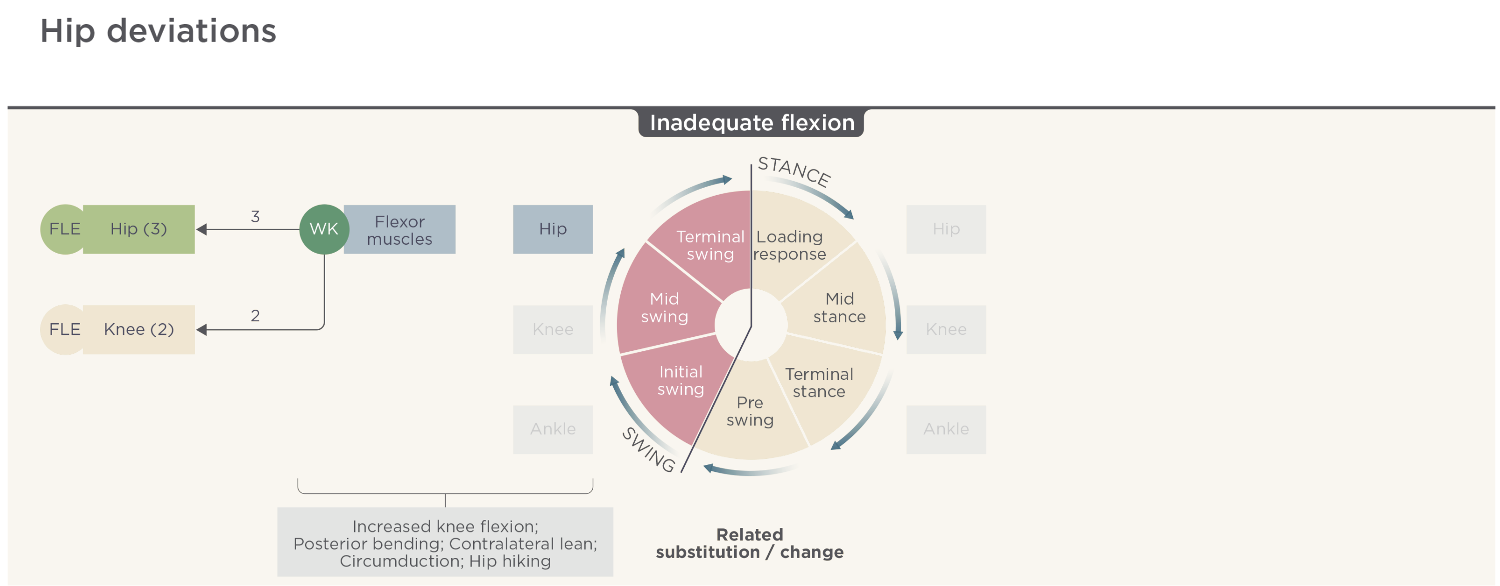


Figure 3.9: Hip joint deviation (inadequate flexion) from normal gait, related deficits and potential support strategies

Ranking of support strategies

The grading scheme produced a ranking of support mechanisms related to their potential to reduce deviations during normal gait at the ankle, hip and knee joint. The evaluation (figure 3.10) has been done for swing and stance phase separately as they are very different from a mechanical and functional perspective (inverted pendulum vs. double pendulum). During stance phase, the majority of ankle joint deviations can be alleviated by external knee moment support; right after support methods that directly target the ankle joint. Maintaining an upright posture without collapsing is a fundamental requirement that permits bipedal gait. If the walking base is altered so that the knee cannot be stabilized with little effort, an external extensor moment can compensate that deviation indirectly. Distal weakness impacts the knee and puts more strain on the quadriceps. The results also show that most knee joint deviations can be mitigated by an external knee extensor moment. Hip deviations are only related to a lack of sufficient extension, and thus, the most beneficial support is an external hip extensor moment. Overall, a knee extensor moment mitigates the deviations by targeting the corresponding deficits directly and indirectly.

The results for swing phase (figure 3.10) show that ankle deviations can be supported best by an external dorsi-flexor moment followed by hip and knee flexion. This comes at no surprise given that the ankle joint mostly needs to keep the foot in a neutral position to ensure ground clearance. Flexion at any of the three joints supports that. An external knee flexion moment is the most beneficial alleviation for knee joint deviations. Others include dorsi-flexor and hip flexor support. The need for external flexor support becomes also apparent at the knee joint. There is only one exception which concerns knee extension. It is needed to straighten the leg at the end of swing phase and prepare the leg for weight-acceptance. External dorsi-flexor support is the best option to mitigate deviations at the hip joint; followed hip and knee flexion. This joint also requires external flexor support. Overall, dorsi-flexor assistance is the best mitigation for swing phase deviations followed by knee and hip flexion. This is in accordance with the fundamental requirement of walking which requires the swinging leg to be able to advance to a position where it can takeover the supporting role. It also serves as an explanation why AFOs that mainly support dorsi-flexion are widely accepted and used.

3.4 Essential requirements derived from pathological gait

Some of the requirements that were identified in the chapter 2 could have also been derived from current devices for patients. These are specifically related to mass distribution and total mass of exoskeletons that provide partial support (RQ1 & RQ2, table 3.1). Motor learning related to efficient usages of the device is also necessary in patient populations (RQ6, table 3.1). Medical devices are a tool. Independently of their intended use in therapy or daily life, exoskeletons depend on the ability of their users. This is especially the case when these devices target certain deficits and provide partial support

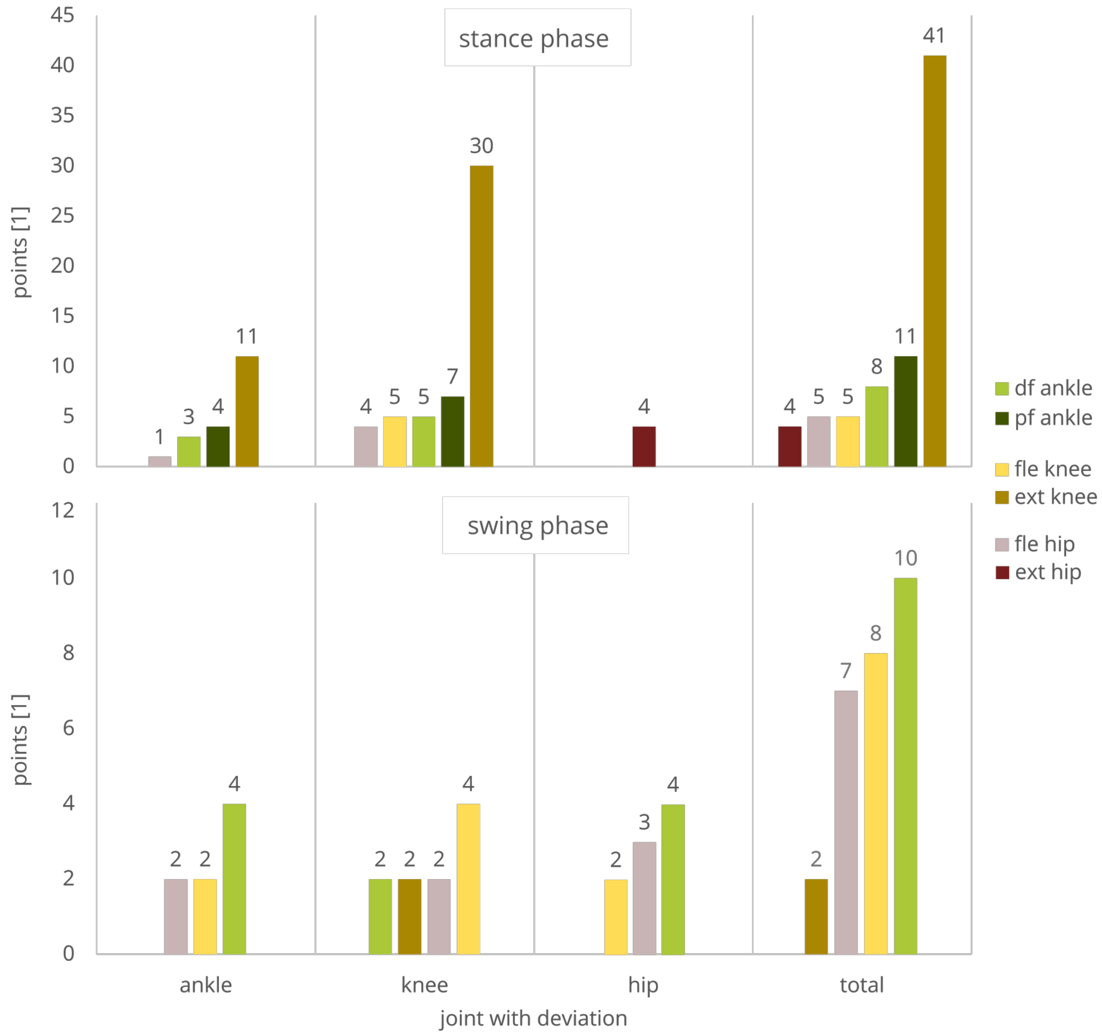


Figure 3.10: Ranking of deficit targeting support methods according to the developed rating scheme.

only. The user has to learn how to operate and use exoskeletons in synchronization with their remaining body function. This principle of learning even applies to passive devices (see chapter 3.2).

New requirements were also derived and can be found in table 3.1 starting from RQ8. Following the deficit based approach, it became apparent that weight-bearing support delivered by an external knee extensor moment is most beneficial. During swing phase, it is favorable to support the flexor muscles at the ankle, knee and hip joint. Weight-bearing is a challenge especially in slow patients [98]. Therefore, it is important to target patients that are classified as household or limited community walkers. If a device can provide assistance that can shift them to a class of higher functioning individuals, it would also increase their level of participation. This is usually one of the main measures for insurance companies to evaluate the cost-benefit ratio of a device or mechanical therapy. Since patients have to learn the use of an exoskeleton, it is important that the

ID	Requirement
RQ1	Light-weight design ($< 6kg$)
RQ2	Minimal distal masses
RQ3	Adjustability to different body shapes
RQ4	Consideration of tissue deformations
RQ5	Optimization of inter-segmental force transmission
RQ6	Intuitive control to facilitate motor learning
RQ7	Avoidance of sensitive pressure points
RQ8	Support of knee extension during stance phase
RQ9	Support for flexor muscles of the hip, knee and ankle joint during swing phase
RQ10	Targeting users that are household or limited community walkers $< 0.8m/s$
RQ11	Structural design allows compensation movements
RQ12	Suitable to be used as therapy and mobility device
RQ13	Partial support in the range of $4Nm - 40Nm$

Table 3.1: Requirements to support and improve pathological movements

device is build and configured to allow free movements without restrictions.

Compensations have usually been learned by patients over many years. It cannot be expected that there will be an immediate transfer to a normal movement pattern just by providing external forces. Hence, an exoskeleton needs to give the user the possibility to "unlearn" their compensation by reducing the compensatory movement little by little. In most cases it will be beneficial if a licensed physiotherapist can support the process. To enable a high dose and intensity of training, it is also advantageous that such a device can be used in therapy and daily life. Given that the presented exoskeletons provide external moments in the range of $4Nm - 40Nm$ and have shown to provide benefits to patients, this range should be also considered for other designs.

Chapter 4

Discussion of essential requirements and design principles

Chapter 2 and 3 derived essential requirements for the design of assistive exoskeletons. This chapter discusses these requirements in more detail and infers designs principles that are later implemented in an assistive exoskeleton: the Myosuit. Table 3.1 summarizes the main requirements that need to be considered when designing an exoskeleton that is designed to partially support the lower limbs and to improve pathological movements. These requirements can be classified into:

- Interface (RQ1,RQ2,RQ3,RQ7,RQ11)
- Provision of force (RQ4,RQ5,RQ8,RQ9,RQ10,RQ13)
- Controls (RQ6,RQ12)

This is not a comprehensive list and only reflects the minimum requirements derived from the findings presented in chapters 2 and 3. The subsequent sections describe each class and the corresponding essential requirements. The implementation of these requirements can be found in detail in the following chapters. Deriving specifications is a standardized technical process, and thus, not considered in detail. That is why this chapter focuses rather on the reasoning behind the requirements and special implications that need to be considered when specifications are derived. The main take-aways, also referred to as design principles, are summarized at the end of each section.

4.1 Interface

Devices that have a positive impact on walking economy distinguish themselves in their low weights and distal masses. This is also a differentiating factor for exoskeletons aiming at improving pathological movements by providing assistive forces. Therefore, it is essential to keep the interface mass low or to distribute the weight so that there is little impact on the person's metabolics [123]. Using passive systems is one potential solution to reduce system complexity and system weight. Alternatively, devices that require distal support move heavy components such as actuators and power supplies usually at a proximal body location. Avoiding cyclic acceleration and deceleration of distal masses can be achieved by either only supporting proximal joints directly or by using remote actuation techniques like cable actuators.

To interface exoskeletons with the human body it usually requires adjustable textiles. Another approach is related to orthotic fabrication techniques that use light-weight carbon fiber or plastic structures. If those are designed to hold and support body parts structurally, the stiff components usually need to be tailor-made for a particular person. Textiles can provide limited structural support, but they are advantageous with respect to their flexibility and adjustability. To avoid slippage of these devices, they must be anchored to the human body. The requirements for the anchoring depend on the resulting forces transmitted to the human body. Normal forces can be easily anchored on any segment. Shear forces, in contrast, require conical shapes for such anchoring. Oftentimes, both forces can be found in exoskeleton interfaces. Ideal body locations that provide such a shape are usually the calf and pelvis. If a direct connection to the upper body does not impose undesireable forces to the upper body, then the shoulders also serve as a suitable anchor. This might be especially important in users where the pelvis does not offer a conical anchor. Body locations that interface directly to an external exoskeleton should also not be sensitive pressure points. These points can be quite individual in patients and depend on the state of the skin (e.g. decubitus) and underlying tissue. Accidents and related changes in body composition and perception might also cause discomfort to a user. The interface should allow for sufficient adjustability to allow the consideration of individual needs.

The interface that connects to the user needs to be made out of inextensible materials to ensure a tight fit and avoid slippage. However, there is a trade-off that needs to be considered when anchoring. Muscles change their volume when they contract. This can be restricted or exacerbated by straps that attach to the muscle. Consequently, blood flow is restricted and the muscles cannot provide energy efficiently which increases metabolic demands [124]. Interfaces should therefore not attach directly to muscles if possible. Nonetheless, this is often impracticable as the exoskeletons support mechanisms need to be aligned with the joints and muscles.

The structure of an exoskeleton should be designed so that it does not impose kinematic changes when unpowered. The previous chapter has demonstrated that motor relearning requires practice with increasing levels of difficulty. To prevent an overwhelming first experience, it is critical that the device can be transparent. Depending on the intended use and related use environments, external boundary conditions need to be taken into account too. This can be related to a use case while sitting or while being in the suspine position. For example, sitting in a wheelchair might restrict the placement of actuators on the back or laterally. Different use cases might impose further restrictions to the system architecture and can also affect the donning and doffing process.
Besides the requirements related to the interface, the Myosuit adheres to the following principles:

- textile based force transmission where possible
- textile based systems to adjust to different users and body shapes (e.g. Velcro

fasteners, buckles, girdles)

- minimally use of rigid structures
- minimal amount of anchor points i.e. pelvis, thigh and shank
- support is provided in the sagittal plane
- lateral movements are not hindered
- design allows to comfortably sit in a wheelchair

4.2 Provision of force

An outcome from the previous chapter is that support of knee extension and the flexor muscles at the hip, knee and ankle joint is key to potentially improve walking function in patients. Chapter 3.2 demonstrated that knee exoskeletons have high potential to improve a patient's kinematic walking patterns. However, the disadvantages of the distal masses limit a successful application. Tendon based approaches have shown that they are advantageous with regard to mass distribution. They can provide assistive forces at distal joints while the actuator is placed proximally. Consequently, a cable actuator can overcome the limitations of current knee exoskeleton designs.

RQ12 requires the exoskeleton to be used as therapy and mobility device. This requirement extends the intended use beyond walking and includes also other motion tasks such as stair ascent and descent as well as sitting transfers that are often trained within physiotherapy sessions. The deviation based approach (chapter 3.3.2) is only able to identify the support method with the highest potential related to walking. However, the method to identify net joint moment synergies (chapter 3.3.1) also considered other movements than walking. Even though this method can only be used with reservations and as a conservative estimation, it reveals the dominant role of extensor activities in movements against gravity. This is in accordance with other investigations that show high muscle activity of the hip and knee extensors at the same time [125, 126, 127, 128]. Therefore, it might be favorable for patients to develop an exoskeleton that can provide assistive extensor force at the hip and knee. A cable actuator can be routed over both joints while keeping the weight of the system low. Chapter 3.3.2 already explained that hip and knee can be supported synchronously from at least $0\% - 28\%$ of normal gait. If the forces are scaled correctly, this could be even extended to pre-swing. This actuation approach is also supported by the theory of the support moment during stance phase [129, 130]. The support moment represents a summation of the extensor moments at all three joints and reflects a consistent extensor pattern. It is less variable than its individual components showing that decreased joint moments at one joint could be compensated by an extensor moment at another. This is especially the case for the hip and knee joint that show considerable variability ($> 60\%$) over stance period, whilst the sum of both shows a reduced variability (21%). Depending on the individual, there is a interrelation between hip and knee that affects the net muscle moments and underlying activation pattern. This indicates the importance to consider force transmissions between joints to generate similar kinematic patterns. Since the support moment has the characteristic double hump shape comparable to the ground reaction forces, it gives an indication of how much the total limb is pushing against the ground to maintain an upright posture. Consequently, it seems beneficial to provide external hip and knee extensor moments for movements against gravity, but also during walking. By supporting two joints with one actuator per leg, we exploit the advantages of cable actuators since the device weight can be kept low. The tendon will be actuated by an electric motor which offers good power density and conrollability, and therefore, enables a force-based control approach.

RQ9 also requires flexor support at all three joints. Adding another tendon would increase the system weight because it would require more actuators or a clutch system. The previous two chapters have shown that passive devices are able to provide a benefit to their users. It has also been demonstrated that passive solutions like AFO can improve the swing phase passively. Similar approaches have been used for providing passive hip flexion support in patient populations [101, 102]. To keep the exoskeleton weight low, the active tendon will be opposed by elastic components (i.e. rubber bands) to provide forces during swing phase at the hip and knee joint. The elastic components will store energy during stance phase which is utilized during swing phase. The opposing moments might also have benefits related to instantaneous joint stiffness and help the body to remain in stable configurations during movements. The ankle joint will be passively held in a neutral position.

The aim is to support patients that can be classified as household or limited community walkers. To track the movement of the hip and knee joint during these slow movements, the tendon speed requirement would be fairly low with $0.45m/s$ (depending on the moment arm configuration of knee and hip joint). However, previous designs have shown that the tissue compression needs to be accounted for as well [131]. The maximum forces can only be delivered if the compliant body interface and the underlying tissue is fully compressed. Initial results of a bi-articular tendon have shown that tissue compression requires up to $20cm$ of travel to generate a force of about $400N$. Taking this into consideration and the fact that the exoskeleton might also need to prevent collapses, the tendon actuator speed requirement is set to $1m/s$. Even though there is a requirement regarding the maximum moments to be generated, it is the required tendon speed that dictates the available forces. This trade-off of force and speed is related to the torque-velocity relation and related dynamics of electric motors.

For the provision of force, the following principles are considered besides the requirements:

- support of hip and knee extensor muscles with one tendon
- active support of stance phase until pre-swing
- use of electric motors
- passive support of flexor muscles (hip,knee) antagonistically
- passive support of dorsi-flexion

- tendon speeds of up to $1m/s$
- actuator placement at the shin due environmental restriction (i.e. wheelchair use)

4.3 Controls

It has been shown in the previous chapters that unimpaired and impaired users require familiarization to benefit from exoskeletons. In order to use an exoskeleton to improve pathological movements, I hypothesize that a consistent exoskeleton-user interaction is key. Patients need to have the possibility to understand the device's behavior to "unlearn" their compensation strategies. This can ideally be done with a licensed physiotherapist that can provide feedback and can also change the force magnitude of the device accordingly. It might be even beneficial to reduce the magnitude of the force to a level that is not noticeable for the user. A posture based support approach that is dependent on a scaling of the knee angle could fulfil that requirement. During movements like stair ascent or standing up, the forces provided are highest when the knee is flexed which corresponds to the extensor muscle activity. This approach can be also implemented from heel strike until pre-swing during walking to stabilize the patient's knee during single leg support. When scaling the force profiles linearly according knee angle during stance phase, the resulting moments are not necessarily linear due to the moment created around adjacent joints by the limb segments. Figure 4.1 depicts the change of the knee moment produced at a constant actuator force of $400N$. The body configuration affects the moment produced considerably. Furthermore, it becomes apparent that the length of the moment arm affects the produced moments mainly while the leg is straight. This behavior needs to be taken into account when designing assistive exoskeletons. The moment arm configuration that we will implemented in the Myosuit is between $4cm$ and $12cm$. The range still provides meaningful support while not extending too far out to become a hazard during use. The moment arm behavior at the hip follows a similar pattern. Because the moment arm length of the hip solely depends on the body composition of the user, generated moments are user specific.

The exoskeleton will use IMUs to implement the posture based control. IMUs represent a small, easily integratable and inexpensive way to measure segment accelerations and angular velocities that can also be used to derive the angle of the lower limbs.

Discussing the implications of the requirements related to the control of the exoskeleton, the following additional principles were derived:

- implementation of posture based force control
- linear force scaling according to knee angle
- set of IMUs to detect events during movements as well deriving lower limb angles (knee,hip,trunk)
- transparent mode without noticeable force application that allows free movements especially during swing phase

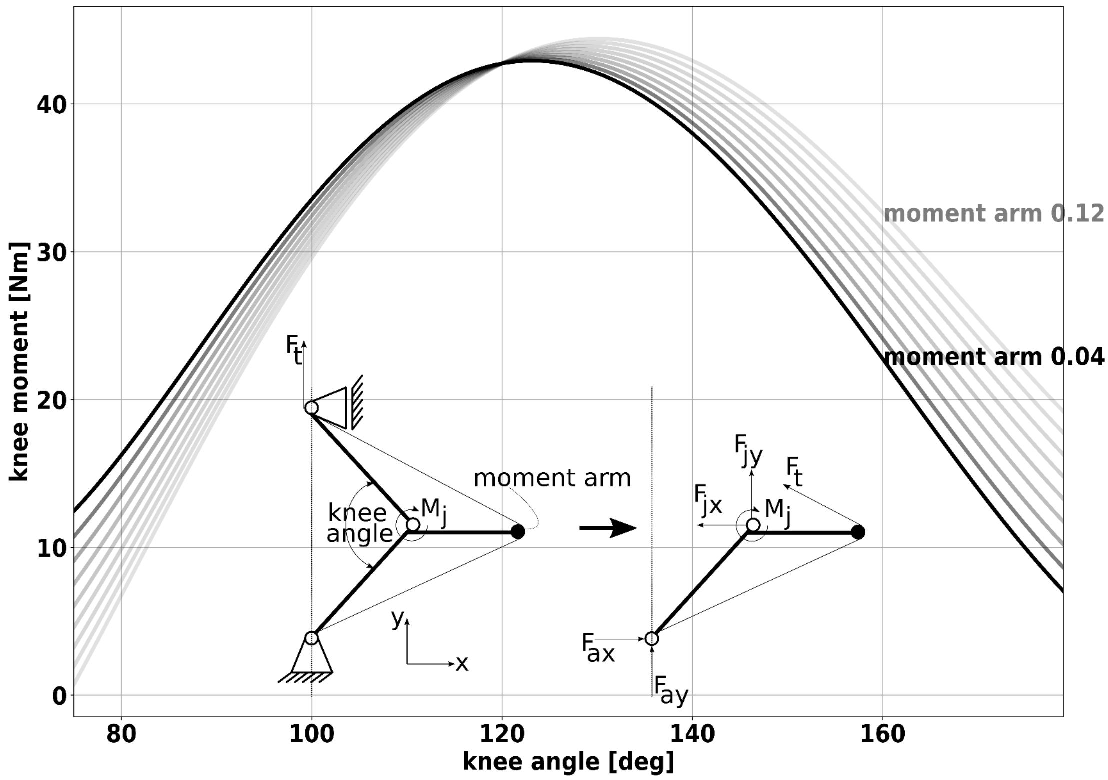


Figure 4.1: Influence of different knee moment arm configurations at a constant output force of $400N$ at the actuator (assuming 40% transmission efficiency).

- adjustability of support levels (force scaling) by the user
- artificial knee moment between $4cm$ and $12cm$
- no artificial moment arm at the hip joint

Chapter 5

The Myosuit

This chapter will present the implementation of the essential requirements and design principles introduced in chapter 4. The developed device is called Myosuit. "Myo" is derived from the Greek word for muscle and "suit" is a reference to the wearability. The Myosuit acts like an external muscle that adds to the net muscle joint moments at the knee, hip and ankle joint in the sagittal plane. The device is designed to use light-weight materials such as textiles and is based on cable transmission. Therefore, the device is also referred to as exosuit. To better understand if the design can provide a benefit to its user, we conducted a first test on one unimpaired participant. We chose to conduct this first test during sitting transfers since the extensor muscles have to provide more energy compared to ground level walking. We expected changes and reductions in muscle activity to be more obvious during this movement.

Cable-driven exoskeletons can be designed in such a way that their actuators encompass multiple joints, mimicking bi-articular muscles which apply forces to adjacent joints at the same time [132]. This means that an exosuit can have a simpler mechanical design by applying forces to two joints using only one motor. Asbeck et al. used a multi-articular strategy to assist hip flexion and ankle planar flexion during ground-level walking [107]. The Myosuit uses yet another bi-articular approach that supports hip and knee extension. In this bi-articular configuration the magnitude of the transmitted moments is strongly dependent on the size of the moment arm across the targeted joints. Based on this concept, and a novel posture-based control approach, the Myosuit delivers continuous assistance to the user when moving against gravity (e.g. getting up from a chair), when moving with gravity (e.g. sitting on a chair) or when holding an upright posture (e.g. standing). We hypothesize that the Myosuit can reduce the muscle activity of knee and hip extensor muscles–specifically gluteus maximus and vastus lateralis–during sitting transfers by providing bi-articular force assistance solely based on the knee angle. We conducted an initial study with a single subject performing sitting transfers and showed that the muscle activity of hip extensors was reduced when the assistive forces were applied.

5.1 The Myosuit concept

The Myosuit is a textile-based robotic device designed to assist people with mobility impairments when performing activities of daily living (ADLs). Most exosuits transmit forces—generated by actuators—to a user through the interaction of garment-like, functional textile anchors and cable-based transmissions [133, 118, 134]. The Myosuit extends this concept by classifying a three-layer architecture, which is inspired by the bones, ligaments, and muscles of the human's musculoskeletal system, to dynamically adapt the levels of support according to the users' needs. The three layers are: a garment layer, a ligament layer, and a power layer (Fig. 5.1). The garment layer is the interface between the Myosuit and the user and provides the overall structure of the suit. The ligament layer incorporates passive elements—rubber bands—to store energy and passively assist the joint's movement. The power layer uses an actuator, routed along the limb, to actively assist the movement of the joint.

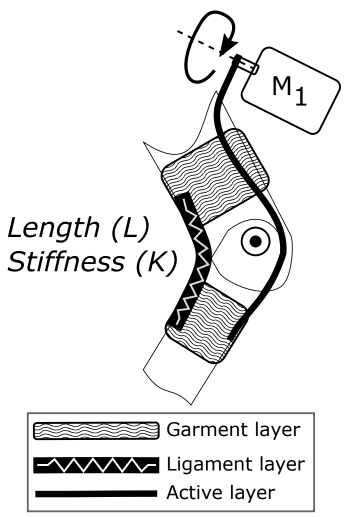


Figure 5.1: Three-layer architecture of the Myosuit.

The garment layer is the interface between the Myosuit and the user and provides the overall structure of the suit and distributes the forces along the body. The ligament layer incorporates passive elements to store energy and passively assist with hip and knee flexion. The power layer uses one actuator and an artificial tendon, routed along the leg from the shank to the waist, to actively assist against the force of gravity at the hip and knee joint (Fig. 5.2). The three layers of the Myosuit have been designed to work similar to an antagonistic pair of muscles to modulate the forces and the stiffness around the biological joints, and thus, provide structural stability in the absence of a rigid frame. The Myosuit's tendon actuators are attached at the shank and anchored to a waist belt at the hip and by a wrap at each foot. The tendons are routed along the leg using textile cable channels sewn to the garment layer. When the support forces are active, the tensile forces translate into extension torques at the hip and knee joint. The ligament layer provides an antagonistic structure to assist with hip and knee flexion.

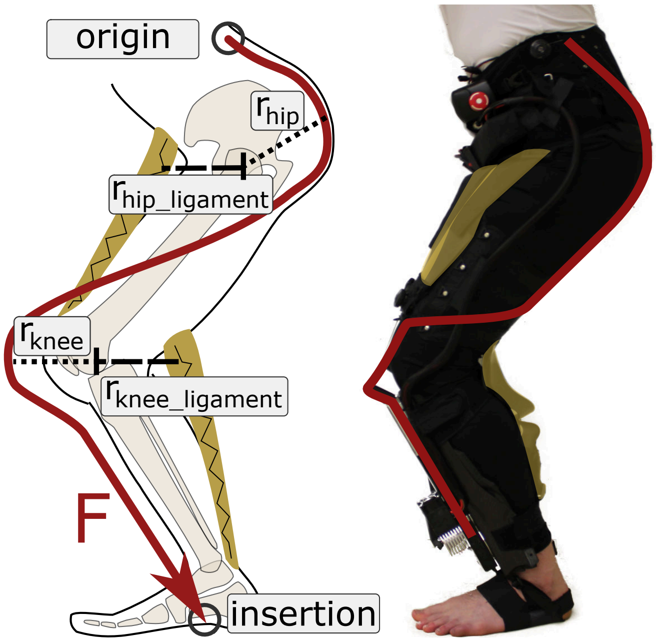


Figure 5.2: Active and ligament layers of the Myosuit.

The Myosuit provides assistance that goes beyond event-triggered bursts of forces relying on specific phases of the gait cycle. Instead, it provides forces continuously either with gravity (damping - eccentric behavior) or against gravity (concentric behavior) throughout the user's movements. The forces are based on the joints' posture, thus resembling the behavior of biological muscles. When the forces in the power layer are scaled down due to decreasing influence of gravity, the transmission of mechanical work of the passive elements in the ligament layer provide support during hip and knee flexion. This force-based approach is made possible by running an embedded, real-time system at 1 kHz that uses closed-loop force control instead of a position based control approach.

The Myosuit combines the assistance aim of most lower limb exoskeletons—anti-gravity support—with the lightweight design of the textile interfaces of exosuits. The total mass of the current system is 4.09 kg without batteries and 4.56 kg including lithium polymer batteries that can power the system up to four hours.

From a functional perspective, the Myosuit must prevent collapse during stance phase and allow the leg to swing freely forward during swing phase. To provide assistance to the hip and knee flexors, the passive ligaments must be able to compensate for portions of the weight of the limb segments and provide support during swing. The importance of the anti-gravity synergy of the hip and knee joint becomes even more apparent during movements where the potential energy of the center of mass (COM) changes significantly. This is for example the case in sitting transfers where the height of the COM changes significantly and the anti-gravity muscles have to provide extensive mechanical work to support the movement. This is why the Myosuit was first evaluated during sit-to-stand and stand-to-sit movements.

5.2 Design

The Myosuit was designed to allow its user to sit comfortably in a wheelchair. Therefore, all rigid components (control unit, batteries, tendon actuators) are distributed across the body without limiting the ability to sit. The control unit and the batteries are mounted close to the COM (see Fig. 5.3) and the tendon actuators are mounted distally at the shank. It is assumed that the COM is in front of the lumbosacral junction when the user is standing upright. The ligament layer compensates for the added distal mass. It is also expected that the user will only move slowly which keeps the influence of inertia low. The Myosuit resembles a pair of pants that include a waist belt and thigh cuffs sewn onto a stretchable base material. One set of rubber bands, designed to aid with hip flexion, attach at the front of the waist belt and the front of the thigh cuffs. A second set of rubber bands, designed to aid with knee flexion, attach at the back of thigh cuffs and the base of the actuation unit. The waist belt and thigh cuffs can be adjusted to the user's body to ensure proper fit of the device and efficient transmission of forces. The waist belt houses the control unit and a set of batteries used to power the actuator units. Power cords connect the control unit to the actuators and provide the required power for the system. Dyneema cables, routed along the garment layer, define the actuation path of the Myosuit.

Figure 5.3: System architecture of the Myosuit

5.2.1 Myosuit layers

The garment layer of the Myosuit resembles a pair of pants (1380 g) and includes a waist belt and thigh cuffs sewn to a stretchable base material (Spandex). It also includes the attachment points for the ligament and the active layer. The integrated waist belt and thigh cuffs (Fig. 5.3) are adjusted to the user's body to ensure a tight fit that minimizes slippage, and thus, allows consistent transmission of force to the biological joints. This is accomplished by a corset-like structure in both the cuffs and the waist belt.

Two Velcro flaps help close the waist belt on top of which the corset applies additional compression. The corset is tensioned on top of multiple polyethylene segments (2 mm thick) that support the lower back and increase the longitudinal stiffness. To minimize shear forces on the skin, the tendons are not attached directly to the polyethylene segments. Instead, a separate layer of nylon and webbing (polyester), placed above the segments, serves as the attachment point. Possible slippage due to tangential forces is limited to the segments. The shearing on top of the segments results in predominantly normal forces transmitted to the user's skin due to the conical shape of the waist. The waist belt prevents the downward migration of the garment layer when the tendon actuators are active.

At the thighs, cuffs equipped with a corset and Boa system (Boa Technology Inc.) help increase the level of compression of the soft tissue and the total stiffness of the Myosuit (Fig. 5.3). Additionally, these cuffs prevent any upward shift of the garment layer and align the tendons relative to the knee. Both the waist belt and thigh cuffs are made of inelastic Cordura-nylon-fabric (330den & 500den), reinforced by layers of polyamide (1 mm) and a Cordura-nylon-Dacron-laminate (X-Pac VX21). The direct interface to the human body is partially cushioned by an inelastic 3D-spacer-mesh (polyester, 3 mm & 6 mm thick).

The active layer uses inelastic and abrasion-resistant tendons (Dyneema (UHMWPE), 1.2 mm diameter) to transmit the assistive forces to the user. The artificial tendons (Fig. 5.3) are guided through the garment layer inside of Teflon (PTFE) tubes connecting the waist belt and the actuators. The tubes avoid small bending radii to reduce friction and use gaps between the tubes to allow for length adjustments of the actuation path i.e. when the Myosuit is active. The tendon actuators, which are part of the active layer, are secured by elastic bands wrapped around the shank and firmly anchored through foot wraps (32 g each) that prevent any upward movement when forces are applied (Fig. 5.3). The actuation path extends from the waist belt (posterior) to the thigh (anterior) until it crosses the knee joint and connects to the tendon actuator (Fig. 5.3). This posterior-anterior transition is done by guiding two artificial tendons symmetrically along the thigh, one lateral and one medial. The routing provides a moment arm at the hip of about 10-12 cm and 10 cm at the knee. The tendons are connected to the actuators by an adapter that helps distribute the forces equally (medial and lateral) when the system is actively pulling. The Boa system at the waist belt (Fig. 5.3) is also part of the active layer and enables the user to adjust the initial tendon length by $\pm$ 10 cm and to pretension the system.

The ligament layer uses passive elements to connect the thigh cuff and the waist belt (anteriorly) and the thigh cuff and the shank (posteriorly) at each leg. The passive elements (rubber bands) are stacks of different Thera-Bands that can be adjusted to achieve different strengths (Fig. 5.3). The current setup uses two layers of Thera-Band Gold and two layers of Thera-Band Black that combine for a force of about 215 N for 30% of elongation. To minimize the shear forces applied to the user's skin, the ligament layer uses the garment layer in a similar way to the power layer. Additional nylon layers and webbing are used to allow possible slippage due to tangential forces on a separate layer on top of the user's skin.

5.2.2 Tendon actuators

Two actuator units, each weighing 1070 g, provide the supportive forces of the Myosuit (Fig. 5.4). The actuator is placed on top of a carbon fiber shin-plate.

Figure 5.4: Tendon actuator weighing 1070 g.

The design incorporates the artificial moment arm (r_{knee} 0.1 m) for the actuation of the knee (r_{knee} in Fig. 5.2). Each unit incorporates a 70 W brushless DC motor (Maxon EC-i40) that is actively cooled by a fan. A Dyneema (0.6 mm) cable is fixed to the motor shaft (6 mm in diameter) and connects via a pulley system to the upper end of the actuator unit. The Dyneema cable is wrapped around the motor shaft multiple times when the system is active. Since the actuators are force controlled, there are no sudden jumps in force due to multiple wraps around the shaft and corresponding change of the effective diameter. The effective diameter cannot be measured or modeled exactly, and thus, there is no accurate position measurement embedded in the unit. It is for this reason that Myosuit's specific characteristics are reported in encoder count dependent values — whereby 2048 counts correspond to 1 rotation of the motor shaft. At the upper end, the actuator cable is connected to the multi-articular tendons connecting to the waist belt. The pulley system transmission ratio is 1:4. The unit allows for a maximum cable travel of 0.24 m, which equals approximately 86000 encoder counts.

The efficiency of the tendon actuators is 86% and the system's efficiency, including the textile layers, is 68%. Therefore, forces at the knee will always be higher than the forces applied to the hip because of the longer cable and increased friction. The efficiency of the tendon actuator was determined by comparing the commanded torque multiplied by angular velocity—the ideal performance—to the force at the tendon attachment adapter multiplied by the tendon velocity. The efficiency of the textile layers was determined in the same way, but on a previous version of the suit which used the same tendon routing and Teflon tubes.

The tendon actuators can deliver 435 N of force during continuous support of ADLs and are able to withstand forces up to 630 N for a short time (for example sudden collapse of the user). The maximum forces depend on the suit's stiffness and the available cable travel; the stiffer the Myosuit layers, the higher the maximum forces for are given cable travel. This is mainly because the maximum cable travel is partially used to compensate slack in the system and to compress soft tissue. The position bandwidth of the tendon actuators is 3.5 Hz. The actuator's bandwidth was tested by attaching it to a subject in a sitting position while performing a sine sweep (range of 0-20cm). For the closed-loop response of the controlled variable, the 3db attenuation from the setpoint occurred at 3.5Hz. The maximum forces achieved during this test were about 435 N. Since the Myosuit is designed for people that are having mobility problems, it is not expected that the suit has to react any faster than traversing almost full cable travel within 280 ms.

5.2.3 Sensors

The Myosuit uses an array of motion sensors to estimate the user's movement intention, movement task, and current posture. Each actuator unit includes a force sensor (FUTEK, LSB200; maximum capacity of 445 N) to measure tendon forces and a motor encoder (Maxon motor, ENX16 EASY; 512 couns/turn) to measure the change in tendon length. The load measures the force in the pulley system before the change in gear ratio of 1:4 (Fig. 5.4). The load cell signal is filtered by low-pass Butterworth filter (2nd order, cut-off 50 Hz, sample rate 1 kHz). There are three inertial measurement units (IMUs), one located in the control unit (STMicroelectronics, LSM9DS1), and two located on each shank (IMU; STMicroelectronics, LSM9DS0). The IMUs provide information on the acceleration (at 0.0012 m/s^2 resolution) and rate of rotation (at 0.0175 deg$/s$ resolution) of the trunk and shanks. The IMU data is transmitted through an I2C bus (400 kHz) and sampled at 100 Hz. A Kalman filter (process noise covariance = 0.001; measurement noise covariance = 0.03) is used to compensate for any drift in the tilt angle calculation for both the control unit near the COM and the shanks.

5.2.4 Control unit

The Myosuit uses two ARM M4 micro controller units (MCU; NXP MK64FN1M0VLL12; 120 MHz, programming language C), one per leg, each running a FreeRTOS Kernel at 1 kHz. The motors are controlled by two servo drives (ELMO, Gold Twitter) that communicate with the MCUs using the CANOpen protocol. The control unit weighs 508 g. The control unit is fixed at the front of the waist belt by Velcro and webbing straps.

Below, two 22.2 V lithium polymer batteries (1200 mAh each, 40C) are fixed using Velcro. The batteries are connected in series for an output voltage of 44.4 V. In case of emergency, the Myosuit can be switched off by an emergency button, located on the side of the control unit, that cuts the power between the batteries and the motors. Other security features include motor speed limits, cable force and length limits, motor temperature limits, and power-down mode (low battery voltage).

A button at the front of the control unit is used to switch between two assistance modes: anti-gravity support—when moving against gravity—and gravity support—when moving with gravity. A second button is used to switch to a transparent mode where the user can move with minimal resistance on the cable (tendon tension 5 N). The system is fully self-contained and does not need any external cable connected during operation. To synchronize the Myosuit with external systems, such as motion capture equipment, an input connection is available to read a triggering signal. Myosuit data (encoder counts, tendon force, raw IMU data, trunk angle, shank angle, synchronization input trigger) of each leg is sent over Bluetooth to a computer at 100 Hz for data logging.

5.3 Closed-Loop control algorithm

The Myosuit provides anti-gravity assistance (with and against gravity) by supporting the extensor torques at the knee and the hip joint. The controller uses an array of motion and forces sensors to continuously estimate the user's movement intentions and posture at 100 Hz. Fig. 5.5 shows the simplified control chart of the control algorithm. The system calculates the influence of gravity on the user's joints—knee and hip—and compensates for this external force by modulating the tendon forces and stiffness accordingly.

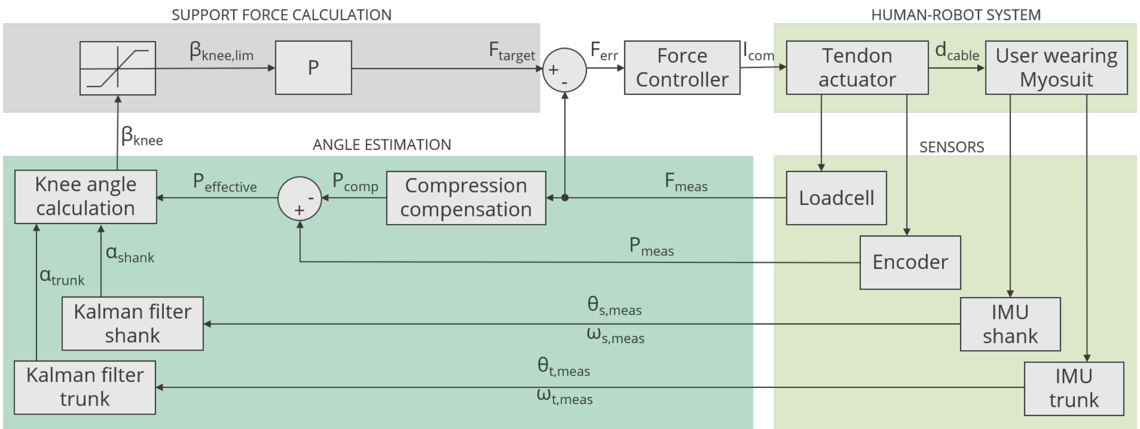


Figure 5.5: Control chart for the Myosuit.

Controller inputs are the shank angle, α_{shank}, trunk angle, α_{trunk}, and the length of the tendon. Using these inputs, the knee angle, β_{knee}, is calculated in real-time and used to adjust the level of force delivered to the user based on the user's current posture—a full derivation is given below. The target force F_{target} is then fed into a PID controller that sets the final force level. The anti-gravity concept can be used with minimal adaptation across the targeted ADLs (ground-level walking, walking up and down slopes, stair ascend and descent, sit-to-stand and stand-to-sit transfers).

5.3.1 Anti-gravity control against the force of gravity

The anti-gravity control provides an assistive extension moment at the knee and hip joints. The controller is based on a virtual leg that connects the hip and ankle joint. The virtual leg length scales as a function of the knee angle β_{knee} (Fig. 5.6). The calculated knee angle is used to adjust the assistive forces dependent on the maximum torque at a knee flexion angle of 80°. As the angle increases, the virtual leg becomes more vertical—and therefore more stable—and the forces is thus reduced.

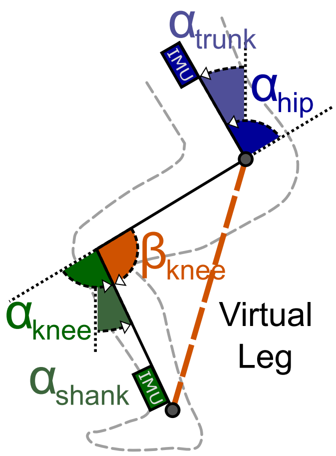


Figure 5.6: Angles measured by the Myosuit.

The knee angle, β_{knee}, is calculated in real-time using the measurements from the system's IMUs and motor encoders. The shank angle, α_{shank}, is determined using the IMU at the actuator unit. The trunk angle, α_{trunk}, is determined using the IMU at the control unit. The beta angle is then defined as the angle between the thigh (direct connection between hip and knee joint) and a vertical line perpendicular to the ground (see Fig. 5.6). The following trigonometric relation exists between these angles:

$$\alpha_{knee} - \alpha_{shank} = \alpha_{hip} - \alpha_{trunk} \tag{5.1}$$

Since both α_{knee} and α_{hip} are unknown, an additional parameter — motor encoder counts (EC) — is used to solve the equation for α_{knee} in equation 5.1. A series of experiments was conducted to characterize the relation between encoder counts and the movements of the knee and hip joints. The relation between encoder counts and the α_{knee} angle, EC_{knee}, was measured by flexing and extending the knee joint from 80 ° to a fully extended position while sitting on an elevated platform. The relation between encoder counts and the α_{hip} angle, EC_{hip}, was measured by leaning the trunk forward against a rail while keeping the legs straight (fully extended knee joint).

To compensate for changing encoder counts due to tissue compression, the Myosuit's stiffness was measured. The suit stiffness is the relation between encoder counts,

EC_{force}, and the tendon force, F_{tendon}, under isometric conditions. For this measurement, the subject remained in a sitting posture — with both the knee and hip angles held constant — while the tendon actuator cycled 23 times from a minimal value of 5 N to a maximum value of 350 N. The measurements were taken while sitting because the system's forces, and therefore the effect of tissue compression, will be highest in this posture. The tendon force of 350 N was chosen because up to this force it was possible for the subject to withstand the forces without any movement.

The change in knee angle (β_{knee}) was calculated using the effective change in encoder counts ($EC_{effective}$). This is the corrected number of encoder counts based on the stiffness of the Myosuit (EC_{force}). The $EC_{effective}$ parameter is defined as the difference between the encoder counts measured by the encoder (EC_{total}) and the estimated encoder counts needed to compress the soft tissue by the applied force (EC_{force}).

$$EC_{effective} = EC_{total} - EC_{force} \tag{5.2}$$

where:

$$EC_{force} = \frac{F_{cable}}{m_{force}}, \tag{5.3}$$

and $EC_{effective}$ is the sum of the encoder counts due to the change of the hip and the knee angle; with m_{force} being the relation between the tendon force and the change in encoder counts (stiffness of the Myosuit).

$$EC_{effective} = EC_{hip} + EC_{knee} \tag{5.4}$$

By replacing EC_{hip} and EC_{knee} in 5.4 we obtain:

$$EC_{effective} = \frac{\alpha_{hip}}{m_{hip}} + \frac{\alpha_{knee}}{m_{knee}} \tag{5.5}$$

Where m_{hip} is the relation between the change in hip angle and encoder counts, and m_{knee} is the relation between the change in knee angle and encoder counts. Finally, equations 5.1 and 5.5 can be combined to calculate β_{knee}:

$$\beta_{knee} = 180 - \frac{m_{knee} \cdot (EC_{effective} \cdot m_{hip} + \alpha_{shank} - \alpha_{trunk})}{m_{hip} + m_{knee}} \tag{5.6}$$

The assistive forces were scaled such that they were lowest in the standing posture ($\beta_{knee} \approx 180°, F_{Tendon} = 5\,N$) and highest in the sitting posture ($\beta_{knee} \approx 100°, F_{Tendon} = 350\,N$). To validate the relation between moment and angle of the hip and knee joint, knee and hip moments were measured during sitting transfers using a motion capture system (Vicon, ten-camera system, Vicon Motion Systems Ltd) and the ground reaction forces using one force plate (Kistler force plate 9260AA, Kistler Holding AG) per leg. The assistance for the sit-to-stand transition was automatically switched on when the tilt angle of the COM exceeded 0 ° during sitting.

5.3.2 Anti-gravity control with the force of gravity

For stand-to-sit transition the anti-gravity control is used similarly to the method introduced before by using the virtual leg calculations β_{knee}. However, the level of assistance was scaled down to allow the subject to sit and not cause locking of the joints; that is, if the force level was the same as during sit-to-stand, the subject was unable to sit and instead had to actively work against the system to flex the knee joint (i.e. contract the hamstrings to overcome the Myosuit's force). A scaling factor of 35% of the sit-to-stand forces was found to work well in pilot tests of the device. A fixed knee angle of 115° was defined to switch off the assistance shortly before chair contact. This was done to prevent large forces from being applied to the user while in the sitting position.

5.4 Experimental protocol

To test the control concept of the Myosuit, a 28-year-old, unimpaired subject (height: 1.8 m; mass: 80 kg) performed ten sit-to-stand and stand-to-sit transitions under non-assisted and assisted conditions. In the non-assisted condition, a constant tendon force of 5 N was set in order to have a taut tendon. In the assisted condition, the tendon forces were defined by the anti-gravity controller previously described. The forces were scaled depending on the transition performed. For sit-to-stand transitions, the scaling value was set to 1; for stand-to-sit, the scaling value was set to 0.35. That is, the tendon forces during stand-to-sit were set to 35% of the sit-to-stand forces.

Between transitions, the subject was instructed to either stand or sit upright. Transitions were initiated by a metronome with 12 seconds intervals. At the beginning of each experiment, the Myosuit, force plates, and the motion capture system were synchronized with a 3V trigger signal. The height of the chair was selected to match the mean wheelchair height of 0.48 m which is equal to DIN 18040-1, the public restroom standard (Fig. 5.7).

Joint kinematics were measured at 200 Hz using the Vicon motion capture system. Ground reaction forces were measured at 1 kHz using one Kistler force plate (Kistler force plate 9260AA, Kistler Holding AG) per leg. Muscle myoelectric activity was measured at 1 kHz for the hip extensor gluteus maximus and the knee extensor vastus lateralis using a wireless Noraxon system (Telemyo DTS). Gluteus maximus and the quadriceps are the main anti-gravity muscles and the main contributors to the sit-to-stand transitions [125]. Since the Myosuit requires a tight fit, it was only possible to measure gluteus maximus and vastus lateralis reliably with the current version of the Myosuit layers. The study was approved by the institutional ethics review board of ETH Zurich. The participant provided written informed consent before participation.

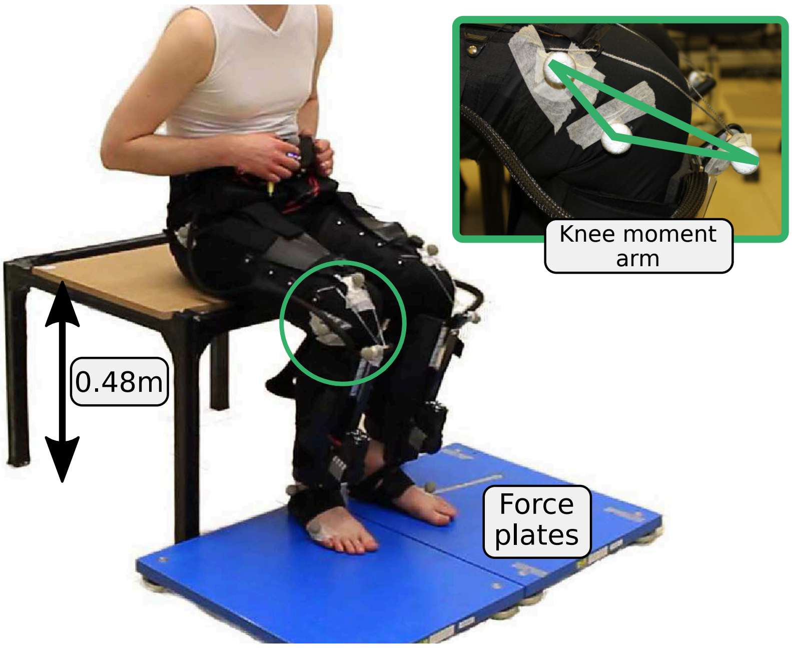


Figure 5.7: Experimental setup for the sitting transitions.

5.5 Data analysis

5.5.1 Kinematics and kinetics

Marker and force data were filtered using a zero-lag Butterworth filter (fourth order). A cutoff frequency of 6 Hz was used for the marker; and 10 Hz for the force data. After calculating joint angles and angular velocities, the kinematic data was filtered offline using a zero-lag Butterworth filter (second-order) with a cutoff frequency of 10 Hz. The hip angle was calculated between a marker placed centered on top of the control unit, the trochanter major, and an assumed center of rotation at the knee (2 cm proximal of the joint space on the lateral femoral condyle). The knee angle was calculated as the angle between the line connecting the greater trochanter major, the rotational knee point and the line connecting the rotational knee point and the lateral malleolus. Angular velocities and accelerations of the joints were calculated by numerical differentiation and were multiplied by the joint moment to obtain joint powers. Joint moments were determined using a static approach [135]. The knee moment of the Myosuit was calculated based on the knee lever-arm and the tendon force. To calculate the knee lever-arm, reflective markers were placed where the tendons exit the Teflon tubes and where the tendon connects to the actuator (Fig. 5.7). The Myosuit knee power was calculated by multiplying the Myosuit moment and the knee angular velocity. Moments were normalized to individual subject body mass.

5.5.2 Muscle activity

EMG data was band-pass filtered (fourth order Butterworth, cut-off 20–450 Hz), rectified, and low-pass filtered (fourth order Butterworth, cut-off 6 Hz). The mean of up to ten repetitions was normalized for each muscle to the total peak value from the four conditions: sit-to-stand, stand-to-sit, both with and without assistance. To reduce the offset, the minimum of the four conditions was identified and it was then subtracted from all conditions.

5.5.3 Transition identification

The sit-to-stand and stand-to-sit transitions were cut and normalized for the subject. To cut and normalize the data it was necessary to identify the point where the gradient of the ground reaction forces reaches its maximum for both transitions (Fig. 5.8). The beginning of sit-to-stand was identified by using the maximum change during the increase in ground reaction forces (sum of both legs). By using previous samples, the time was identified were hip angular velocity and hip angular acceleration became more negative than -10 deg/s and -10 deg/s^2 (sum of both legs), respectively.

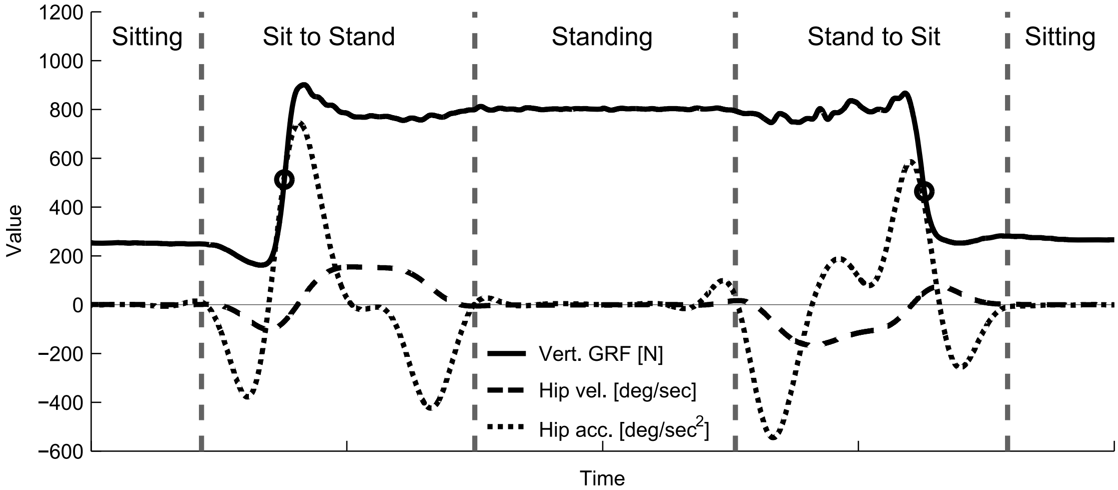


Figure 5.8: Example hip kinematics and ground reaction forces used to segment the recorded data offline for the analysis.

The end of the sit-to-stand transition was identified by using the next samples to find the time when the hip angular acceleration became more negative than -10 deg/s^2 and the hip angular velocity fell below 10 deg/s. The stand-to-sit transition was identified using the maximum change during decreasing ground reaction forces following the same method explained above. The time was normalized for the first part of each transition (from start to the point of the maximum force gradient). In addition, the second part was normalized from the point of maximum change in force until the end of the transition. The length of the first and the second part was set by determining the average time for the first and second part for all transitions before time normalization. This method to identify transition was used offline after data collection.

5.6 Results of the first testing

5.6.1 Compression compensation

For the relations between joint angles and motor rotations we report the more standard measure of angles: radians. Note that our internal calculation relied on the encoder counts (EC) as described in the Methods section of the article. The relation is given by: $1EC = 2\pi/2048EC$. We acknowledge that EC is not a standard engineering unit nor a hardware independent variable. However, it is important to point out that the tendon (cable) is wrapped around the motor without being guided. This means the change in winch diameter due to multiple cable wraps is not known, and thus, the exact relation between tendon length and motor rotation is not known either. Since the system does not rely on internal cable position measurement and purely on the tendon force, this design does not affect the usability of the system.

The relation between the change in knee angle and encoder counts (m_{knee}) was approximated as linear with a slope of: 0.75 deg/rad (Fig. 5.9a; bottom row). The relation between the change in hip angle and encoder counts (m_{hip}) was approximated as linear with a slope of: 0.68 deg/rad (Fig. 5.9b; bottom row). The relation between the force on the tendon and the change in encoder counts (stiffness of the Myosuit), under the isometric condition, was approximated as linear with a slope of: $EC_{force} = 5.28$ N/rad (Fig. 5.9c; bottom row) which approximately corresponds to 5400 N/m. This curve shows a distinct hysteresis that the linearization of the loading phase cannot account for. These three relations were used to determine the knee angle β_{knee} in real-time using the virtual leg controller. However, it was found that this simplification still showed acceptable estimates for β_{knee}.

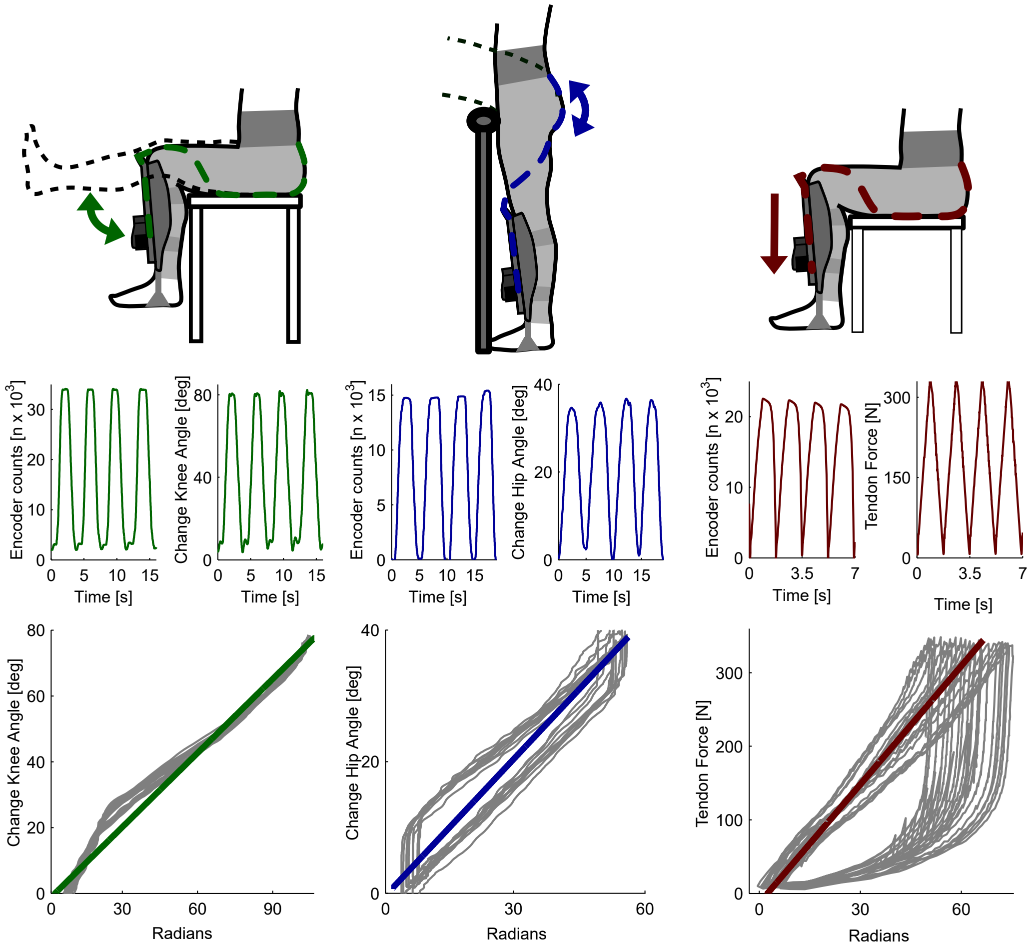
For the relation between knee and hip moments and knee and hip angles, a linear relation between angles and moments from 90° to 170° was obtained (Fig. 5.10). The curves were experimentally determined. This linear relationship was approximately the same for transfers with and against gravity. Scaling the forces linearly dependent on the virtual leg will apply considerable and biological relevant torques to both the hip and knee joint. Since the Myosuit does not provide 100% of the biological torque, the forces were scaled linearly to the maximum continuous force during ADLs (435 N) and the lowest possible force (5 N; non-assist mode).

The range of angles chosen (80° to 180°) to scale the assistive forces was slightly bigger then the one measured in Fig.5.10. This is the maximum range a user is expected to use in the suit during everyday life.

The following linear equation describes this behavior accounting for the artificial moment arm $r_{knee} = 0.1$ (Fig. 5.2):

$$Force = \frac{-0.43125 \cdot \beta_{knee} + 78.125}{0.1} \tag{5.7}$$

This relation is used to scale the assistive forces of the Myosuit.



(a) $m_{knee} = 0.75\,deg/rad$ (b) $m_{hip} = 0.68\,deg/rad$ (c) $EC_{force} = 5.28\,N/rad$

Figure 5.9: Myosuit characterization.

5.6.2 Posture based anti-gravity control

The estimated knee angle, based on the virtual leg model, was used successfully to grade the level of gravity support provided to the user (Fig. 5.11). During sit-to-stand, the estimated knee angle (β_{knee}) lagged the measurement from the motion capture system by an average of 11.48% at the onset of the movement. This lag decreased to zero as the subject reached the standing posture (Fig. 5.11A). During stand-to-sit, the estimated β_{knee} lagged the motion capture measure by 4.1%; this lag was present through the movement. The knee angle, as estimated by the Myosuit, had a minimum set to 110° due to a safety measure designed to prevent the controller from applying large forces while the subject was in the seated position. At this angle, only pre-tension forces of around 30 N were acting on the subject.

The performance of the controller was characterized by its ability to track the desired tendon force (Fig. 5.11B). While the subject was in a sitting position, the tendon force properly tracked the desired pre-tension force of 30 N. Once the sit-to-stand movement started, the peak tendon force lagged the peak reference force by 11.2% and was lower by

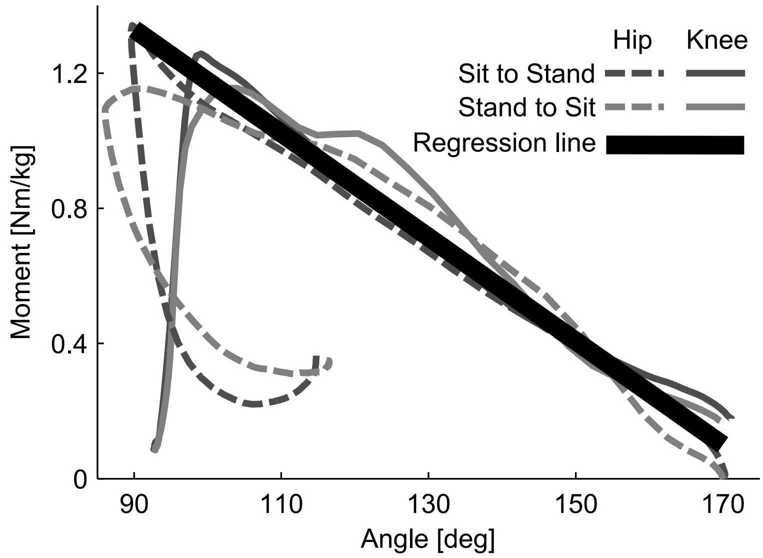


Figure 5.10: Moment-angle curve of hip and knee joints during sitting transfers.

35.72 N. During standing, the force on the tendon was tracked properly at the minimum value set to maintain the tendon taut: 5 N. During stand-to-sit, the largest difference in tendon force occurred towards the end of the movement—close to the sitting position when the forces were the largest—and was, on average, higher by 9.84 N. The transition times that correspond to 0-100% of the movement varied from 1.5 s to 1.9 s.

5.6.3 Joint kinematics and kinetics

During sit-to-stand, the Myosuit delivered about 26% of the peak moment (0.27 Nm/kg) in the assisted condition relative to the non-assisted condition (1.04 Nm/kg) (Fig. 5.11C). For stand-to-sit transitions a peak moment of about 0.1 Nm/kg was provided. This corresponds to about 11% of the peak total moment of the transparent condition (0.9 Nm/kg). Prior to the onset of the sit-to-stand transition, there was a time-related offset between the non-assisted and assisted conditions of 3.9%. During the stand-to-sit transition, there was an increase of the total knee moment in the assisted condition as the subject reached the sitting position.

The Myosuit provided about 35% of the knee peak power of the transparent mode (0.4 W/kg of 1.14 W/kg) during the sit-to-stand transition (Fig.5.11D). For the stand-to-sit transfer, about 15% of the peak power was provided by the Myosuit in the assisted condition when compared to the non-assisted condition (-0.13 W/kg of -0.89 W/kg). Peak power of the assisted condition in the stand-to-sit transfer was higher when compared to the unassisted condition.

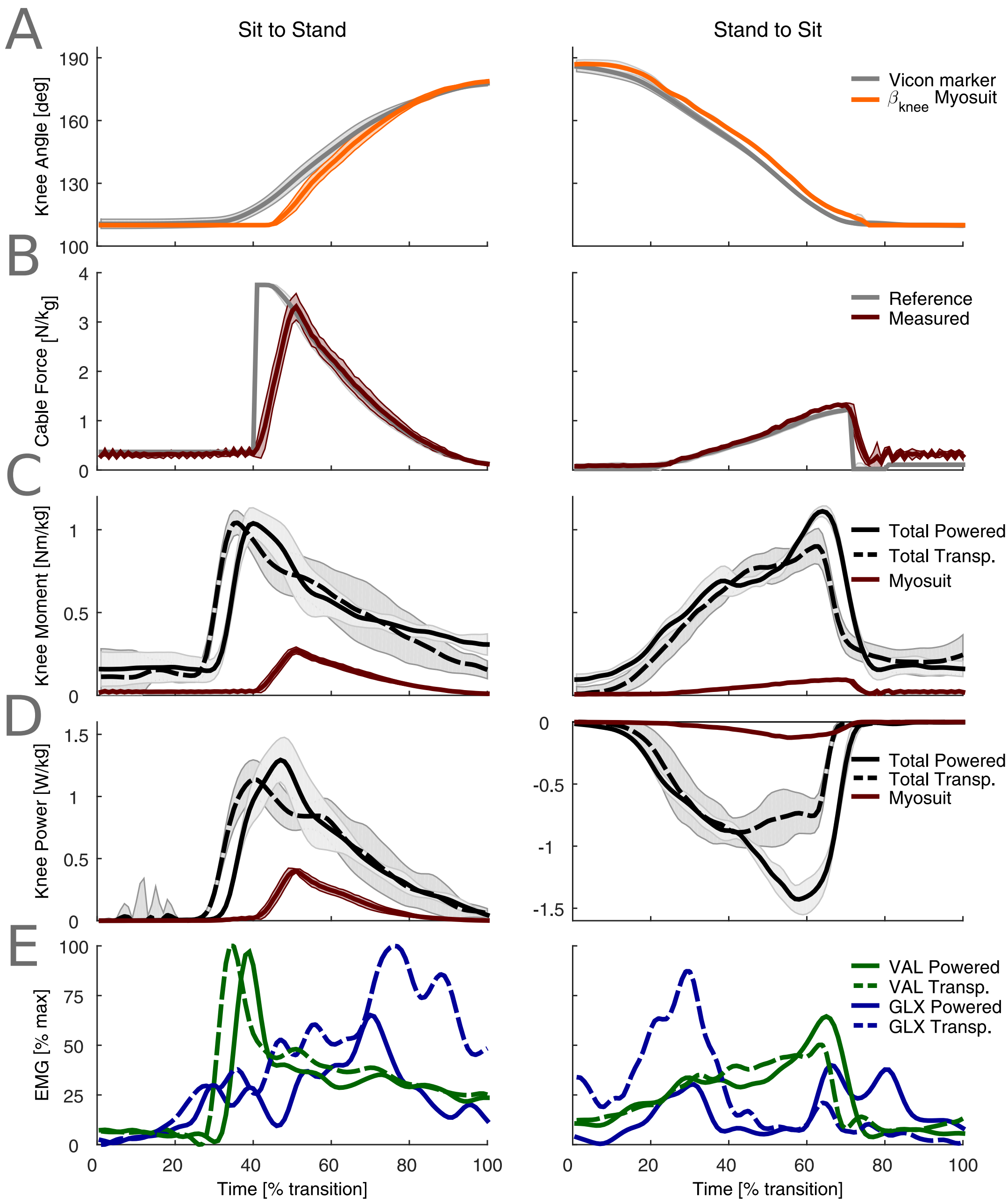


Figure 5.11: Experimental evaluation of the Myosuit during sit-to-stand (left) and stand-to-sit (right) transitions.

5.6.4 Muscle activity

For both sitting transfers, there was a general decrease in the muscle activation of the gluteus maximus (GLX). This reduction was present across the sit-to-stand transition, especially towards the end of the sit-to-stand transition. For the stand-to-sit transition, the reduction was largest at 30% of the transition where it was reduced by about 60%. Two additional muscle activity peaks in the gluteus maximus were present during assistance, at about 60% to 90% of the transition. These peaks were not present in the non-assisted condition. For the vastus lateralis (VAL), there were no clear differences in the magnitude of the muscle activity between the two conditions. During sit-to-stand, there was a lag in the onset of muscle activity, for both the gluteus maximus (GLX) and vastus lateralis (VAL), during the assisted condition (Fig.5.11E).

5.7 Discussion of first testing results

The Myosuit is a lightweight and untethered wearable robotic device that uses a bi-articular design to counteract gravity by actively supporting hip and knee extension. The Myosuit aims at reducing the user's physical effort, quantified in this study by the changes in muscle activity of the knee extensor vastus lateralis and the hip extensor gluteus maximus. Instead of relying on predefined position-based triggers, the assistance concept presented uses a force-based approach that is related to the user's current posture. An initial test showed that the proposed control approach properly estimates the user's posture and intention in sitting transfers. Specifically the knee angle could be estimated by using the exosuit's tendons and linearized suit stiffness. Based on the knee angle, the Myosuit delivered positive and negative power to the user, resulting in up to 26% of the natural knee moment and up to 35% of the knee power. As a result, muscle activity of the hip extensor (gluteus maximus) was reduced in both sit-to-stand and stand-to-sit transfers. The Myosuit's novel approach to assistance using a bi-articular architecture, in combination with the posture-based force controller, can effectively assist its users in gravity-intensive ADLs, such as sitting transfers.

5.7.1 Compression compensation and knee angle estimation

A key component of working with a soft, textile-based structure is the optimization of the transmission of power from the robot to the user. For the Myosuit, the main inefficiencies arise from the friction between the tendons and the garment layer and from the compression of the user's tissue due to the tendon's tension. These effects are illustrated in Fig. 5.9c where the motor winds close to 6 cm of tendon ($\approx 20,000$ encoder counts) to reach 324 N in an isometric condition. While in the loading phase the behavior is primarily linear, the unloading phase presents a clear non-linear relation between the force on the tendon and the winding of the motor. This unloading phase exhibits hysteresis and is characterized by decompression of the user's tissue and by the presence of friction between the tendon and the garment layer.

An unexpected finding was that only the loading phase of the curve had to be taken into account to correct for the Myosuit's stiffness. During initial tests, both the loading and unloading phases of the curve were taken into account when correcting the encoder counts. However, this led to an offset in the encoder counts — and an error in the estimation of β_{knee} — during actual movements. Instead, if only the loading phase of the curve was used, the estimation of β_{knee} was far more accurate. This is likely due to the increased stiffness of the muscles when a movement is executed, as opposed to the isometric condition used for characterization where the muscles were inactive.

The ligament layer can also affect the transmission of joint torques to the user. Although it was inactive in this study — sitting transfers do not require assistance of hip or knee flexion —, other activities of daily life (e.g walking) involve active flexion of the hip and knee. By engaging the ligament layer, the maximum joint torques in extension will be reduced. That is:

$$torque_{net} = torque_{activeLayer} - torque_{ligamentLayer}; \tag{5.8}$$

where the $torque_{ligamentLayer}$ is dependent on the joint's posture and the moment arm of the ligament layer $r_{joint\_ligament}$ (see Fig. 5.2). Fortunately, the decrease in net torque is accompanied by an increase in joint stiffness, similar to a co-contraction of antagonistic muscles, which translates into increased stability for the user.

5.7.2 Posture based anti-gravity control

In a first experiment, sit-to-stand and stand-to-sit transitions were successfully assisted by the anti-gravity controller used by the Myosuit. The assistance concept uses a bi-articular structure to assist with knee and hip extension. The controller was able to identify the knee angle and scale the assistive forces accordingly. Compared to the change of angle detected by the motion capture system, the Myosuit's estimate showed a lag in detection. This might be due to inability of the system to detect changes in cable length without assistance because of its compliant structure. For safety reasons the minimal knee angle was set to 110° for the experiment. The lag could also be due to the fact that the system did not calculate the joint angle correctly and estimated a smaller angle than 110°. The anti-gravity support was triggered if a specific COM angle was exceeded. This simple trigger to switch on the anti-gravity assistance may have been set too late. A clear delay can be seen in the moment, power, and EMG plots. This needs to be validated in a further study.

The assistance applied resulted in 26% of the biological peak of knee torque during sit-to-stand transfer. The initial scaling of the maximum force (435 N) was set for an angle of 80°. In this case the minimum angle was 110° which resulted in a lower desired force (about 306 N). In addition to the late onset of force, the ramping up of force by the tendon actuator was set too low; this led to the lag in force relative to the desired setpoint. In this test, this lag led to a significant reduction in the maximum peak force. Nevertheless, the Myosuit was able to deliver about 35% of the peak power during sit-to-stand movements.

The power curves show an increase in power compared to the non-assisted condition. This is likely due to the increase in speed while standing up. While the vastus lateralis did not show a reduction in muscle activity, the peak EMG activity occurred at the onset of assistance. At this time, however, the Myosuit was not applying significant forces to support the movement. Applying the assistive forces earlier in the transition will likely lead to reduction of the muscle activity at the vastus lateralis.

On the other hand, the muscle activity of the gluteus maximus was reduced during the assisted condition reduction in the sit-to-stand transfer. This indicated that the forces, which are based on the knee, and not the hip angle, apply assistive forces that are biologically relevant and can assist the function of the hip extensors.

The applied forces resulted in 11% of the biological peak knee torque and 15% of the peak knee power during stand-to-sit transfer. The angle estimation shows a slight delay, but it is able to track the change of the angle reliably. An increase in peak moment and negative peak power can be seen before the Myosuit changes in the non-assist mode. This happened before the user touched the chair since the assistance was turned off based on a fixed angle. The sudden release of the force led to the user performing a movement that was less controlled when sitting; this effect likely affected some of the results. For example, due to a harder landing on the chair, the EMG of the gluteus maximus muscle seemed to have increased at 60% to 90% of the transition. This effect on the EMG reading might be a movement artifact caused by the less-controlled landing on the chair.

Immediately before the assistance was switched off, there was an increase in the EMG signal of the vastus lateralis. This was likely the result of a behavioral or learned muscle activation since the user performed multiple repetitions and the assistance was always turned off at the same knee angle. Thus the user may have developed this strategy to prevent a hard impact on the chair. To improve the stand-to-sit transition, the anti-gravity control should be changed from using a constant factor for scaling and instead include a velocity-dependent damping factor. However, a reduction in muscle activity of the gluteus maximus can be observed again, indicating that the applied forces are assisting with the hip extensors.

Chapter 6

Separating Gait Phases

The previous chapter introduced the main architecture of the Myosuit in accordance with the essential requirements and design principles established in chapter 4. The control of the active tendon of the Myosuit is based on the scaling of the knee angle. The tendon needs to apply a force to support weight-bearing when the leg is in stance phase and needs to switch to a transparency mode when the ligaments support the swinging leg passively. To implement such a strategy, we had to develop a gait phase detection algorithm. In order to evaluate different detection options, we have used a measurement setup that is independant of the Myosuit. However, this particular setup can easily be integrated in the Myosuit. The initial tests have been conducted on unimpaired participants to reliably compare the different options. Patients would not provide a consistent basis for such an evaluation. Furthermore, it is theorized that dominant features that reflect the fundamentals of walking ability are also detectable in patients.

6.1 Segmentation options

Walking is a cyclic process with a recurring pattern to move in space while maintaining static and dynamic balance [2]. One cycle of walking, also known as a stride, starts and ends with consecutive heel-strikes of the same foot. A separation of the stride into stance and swing phases is widely used to differentiate when the foot is in contact with the ground and when the foot is in the air [136]. Powered assistive devices use knowledge of the movement phase to tailor their assistance. These devices can be used by people with mobility impairments to train during rehabilitation [137, 138] or to provide assistance in their everyday life [118, 139, 140, 141, 142, 122]. One of the major movements to assist with lower limb exoskeleton is walking. Walking is a cyclic process that comprises multiple reoccurring patterns to move in space while maintaining static and dynamic balance [2]. A proper characterization of the walking phases is crucial since incorrect timing can increase user effort and the chance of stumbling or falling.

Powered assistive devices rely on sensors to detect the phase of gait of the user at a given time. Pressure insoles [143, 144, 145] and foot switches [146, 147, 148] are commonly used because they provide reliable signals related to the loading of the foot. Signals from accelerometers [149, 150, 151], gyroscopes [152, 153, 154, 155, 156] and electromyographic (EMG) sensors [157, 158, 159] can also provide signals for gait segmentation. Capacitive sensors [160] and ultrasonic sensor arrays [161] can also be used to measure the shape of the user's muscles when the limb is inside of a cuff, but they are less common due to their added complexity. In a typical use case, sensor signals are fused together to increase the signal quality by reducing the uncertainty compared to when one sensor is used alone. A common example of fusing sensor signals to segment gait is the use of inertial measurement units (IMUs) [162, 122, 163] which consist of accelerometers, gyroscope, and magnetometers. Another approach is to combine gyroscopes with pressure insoles [164] or gyroscopes with force sensors [133].

The algorithms for phase detection usually consist of state-machines with simple thresholding and peak detection [164, 152, 162, 133], or with more advanced techniques such as oscillator-based algorithms [165], Fuzzy-logic [166, 167], Bayesian inference [168], Hidden-Markov models [148, 155, 169, 170], or neural networks [171, 172]. Advanced algorithms tend to be more versatile for detecting different walking patterns and conditions. However, their implementation in microcontrollers, as they are most commonly used in exoskeletons, is challenging. This is critical if such algorithms are meant to be used in certified devices that reach the market, where regulatory bodies must ensure safety in all circumstances. While state machines allow for deterministic systems where all possible outcomes can be examined, more advanced algorithms often rely on probabilistic outputs. The certification of probabilistic algorithms is difficult since, by nature, the state of a system cannot be known at all times. Most devices currently in the market thus rely on state machines. This article focuses on the suitable implementation for deterministic state machines.

Gait segmentation based on kinematics of the lower limb has achieved good results in previous work [152, 162]. Instead of relying on the biological lower limb segments alone, Villarreal et al. extended the concept by creating virtual limb segments of the lower limb based on geometrical projections of the thigh and shank [173]. The kinematics of the virtual limb segments showed improved monotonic behavior during stance and swing phase compared to the biological segments. A change in the angular direction of a limb segment angle reflects a change in sign or a zero crossing of the corresponding angular velocity.

Vilarreal et al. used the defined lower limb segments as time-independent phase variables to segment gait. These virtual limb segments also show great potential for event detection in a time-dependent manner based only on their angular velocity.

We hypothesized that changes in the sign of the angular velocity of lower limb segments, as shown in figure 6.1, can be used to identify stance and swing phases during walking at different speeds and inclinations. In addition, we hypothesize that the timing of the sign changes can be used to identify gait events such as heel-strike and toe-off. In this article, we studied how the kinematics of biological and virtual lower limb segments can be used for gait segmentation. We first analyzed the suitability of four lower limb segment models (biological and virtual) by determining the sign changes of the angular velocities based on optical motion capture data. A suitable model should only have two zero crossings: one near heel-strike from negative to positive, and one near toe-off from positive to negative (Fig. 6.1). The four models use different kinematic configurations

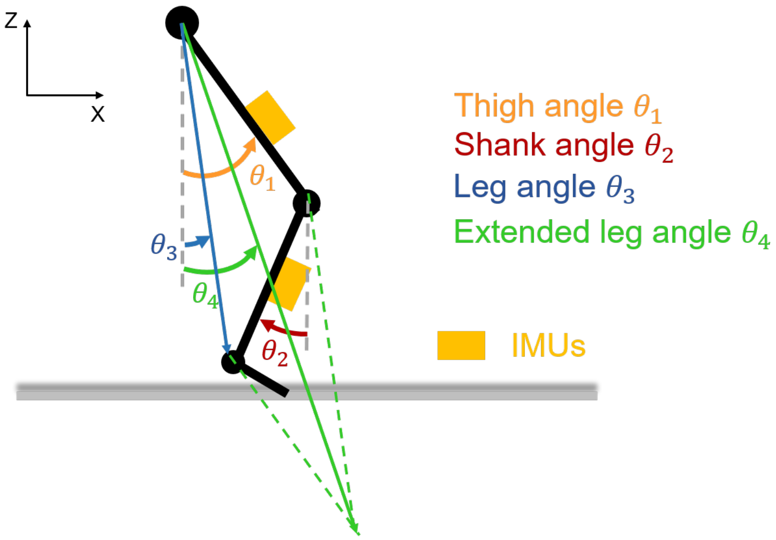


Figure 6.1: Lower limb segment models evaluated for the potential on stance and swing detection.

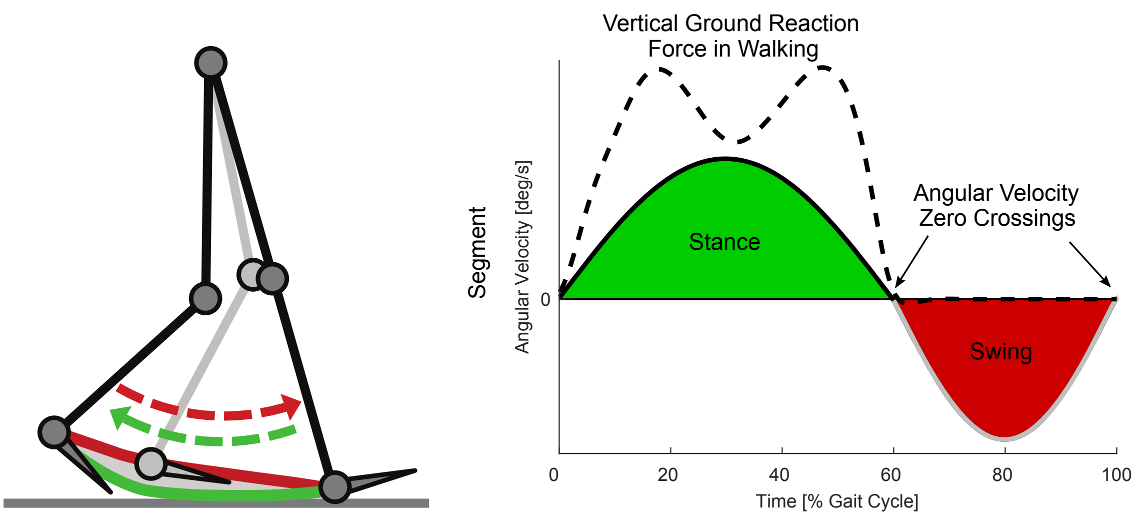


Figure 6.2: Lower limb segment model velocity based concept for stance and swing detection.

of biological (thigh and shank) and virtual (leg and extended leg) lower limb segments. We analyzed the reliability of these kinematic configuration to differentiate between the stance and swing phase. We further analyzed the accuracy of detecting key events such as heel-strike and toe-off. Finally, we evaluated whether a simple sensor arrangement of IMUs could provide good signals for gait segmentation and detection of gait events.

6.2 Study protocol

We recorded the walking kinematics of 13 subjects (5 female and 8 male, age 26±4 years, height 178±11 cm, weight 69.5±11.5 kg). Subjects walked on various ground inclinations and at various walking speeds on an instrumented split-belt treadmill (R-Mill, Forcelink, Netherlands). Walking speeds during level walking were 0.5, 0.9, 1.3, 1.7, and 2.1 m/s. Walking speeds during decline and incline walking were 0.9, 1.3, and 1.7 m/s. The decline and incline angles where -10, -5, +5, and +10 degrees. The preferred walking speed of younger adults is 1.3 m/s [174]. The lowest walking speed (0.5 m/s) was selected based on reports from the subjects that they could walk normally. The fastest walking speed (2.1 m/s) is the preferred transition speed to running [175].

Ground reaction forces (GRF) were recorded using the instrumented treadmill. The kinematics of the lower limb segments were recorded simultaneously using a motion capture system (Vicon Motion Systems, UK) and a set of five inertial measurement units (9-axis IMUs; MPU-9250, TDK, Japan). The IMUs were packaged in the SimpleLink SensorTag (CC2650STK, Texas Instruments, USA) and Bluetooth Low Energy was used to transmit the IMU data to a Single Board Computer (RPi3, Raspberry Pi Foundation, UK). All systems were synchronized before the data collection.

Subjects first walked on level ground, then uphill (5° and 10°), and downhill (5° and 10°). The protocol required subjects to walk continuously in each inclination condition. They rested for five minutes between level, uphill, and downhill walking. Breaks of less than one minute were required to change between slopes from 5° to 10°. We recorded one minute of continuous walking for each walking speed, starting with the slowest speed. In our analysis, we focused on the highest slope (10°), and the speeds 0.5, 1.3 and 2.1 m/s for level walking. All model calculations were performed offline.

6.3 Kinematics and ground reaction forces

Three-dimensional ground reaction forces were recorded at 1000 Hz. The vertical GRF was used to determine the reference times for the heel-strike and take-off of each stride. A vertical GRF greater than 400 N was used as the starting point to identify the upcoming heel-strike event. An algorithm scanned for the first sample to fall below 0 N or that is greater than the subsequent force value and less than 10 N. To determine toe-off time, the GRF were filtered with a zero-lag, second-order, low-pass Butterworth filter (cutoff frequency 20 Hz). Starting 100 ms after each heel-strike an algorithm identified the toe-off time as the time when the GRF value fell below 10 N.

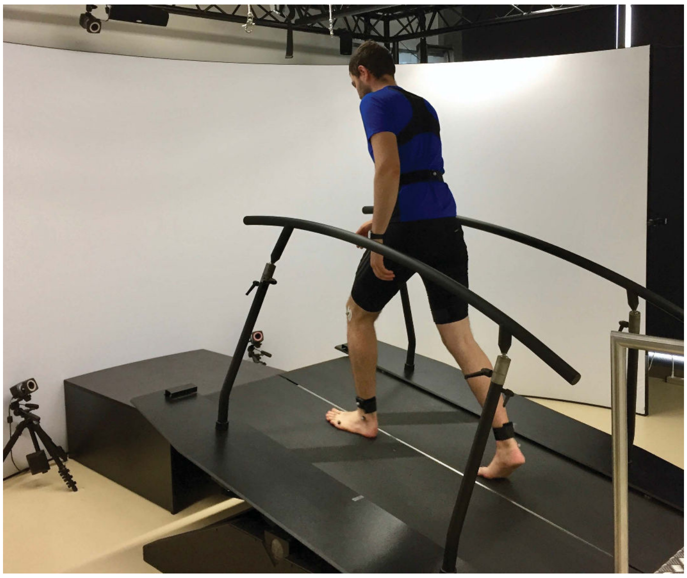


Figure 6.3: Measurement setup including an instrumented split-belt treadmill and a motion capture system.

Body segment kinematics were recorded at 200 Hz using reflective markers from the motion capture system. The marker data was up-sampled linearly to 1000 Hz to match the frequency of the ground reaction forces. Raw marker data was filtered with a zero lag, fourth-order, low-pass Butterworth filter (cutoff frequency 10 Hz). The thigh and shank angles, as well as the angular velocities, were filtered using the same method. The thigh segment was determined by a marker at each hip (greater trochanter head) and each knee (2 cm proximal of the joint space on the lateral femoral condyle). The shank angle was determined by the knee marker and an ankle marker (lateral malleolus). The leg angle and extended leg angle were calculated as described in [173]. The angular velocities of the leg and extended leg model were determined by numerical differentiation.

In addition, kinematics of the thigh and the shank of each limb were recorded using IMUs (Fig. 6.4). The sensor units were placed in textile straps. The length of the straps were adapted to differences in body composition using velcro. Reflective markers on the sensors were not used for this evaluation. Thigh and shank angles were determined by combining gyro and accelerations signals using a Kalman filter (process noise covariance = 0.00005; measurement noise covariance = 7). When calculating the leg and extended leg angle [173], thigh and shank length were assumed to be equal (24.5% body height, based on [2]).

The average gait kinematics of each subject were determined based on 65 to 160 strides (sum of both limbs) for each condition (slope, speed). The average data of each subject was aggregated to obtain the average kinematics of all subjects. The data analyzed included 10737 strides for walking uphill, 12246 for walking downhill, and 13131 for level walking.

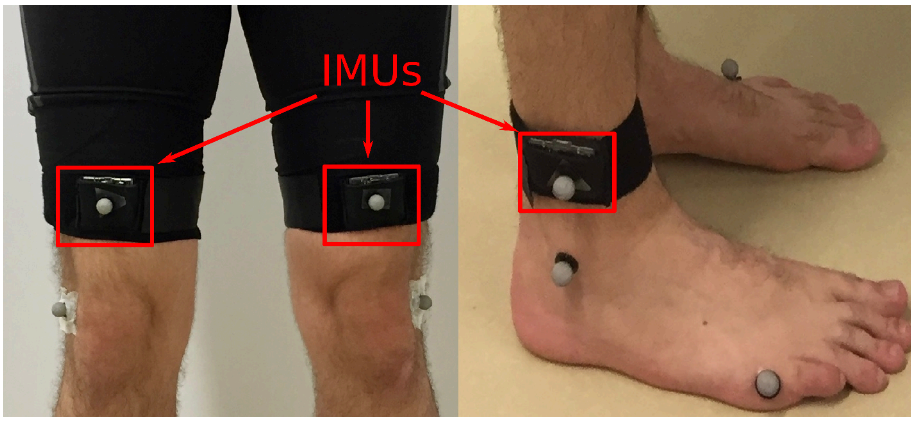


Figure 6.4: Inertial measurement unit (IMU) setup for the thigh and the shank.

6.3.1 Sign changes of the angular velocity

The number of zero crossings of the angular velocity for each lower limb segment model was calculated by aggregating all transitions per stride. Since sign changes are to be used to indicate the transition between stance and swing phases, only one transition of the angular velocity from positive to negative and one from negative to positive were biomechanically relevant (Fig. 6.1). Additional transitions resulted in higher counts (mean value more than one) of zero crossings. Each transition was evaluated with respect to the reference event (heel-strike or toe-off) identified using the vertical GRF. The distance to the reference event was calculated in percentage of the gait cycle. A detection before the reference event results in a negative percentage whereas a detection after the reference event results in a positive percentage.

6.3.2 Constraint rule sets for event detection

Detection constraints were designed to increase the likelihood of detecting correct events by using temporal and causal relations as well as absolute thresholds. The constraints where tailored to each lower limb segment model to detect only one transition in angular velocity from negative to positive and one from positive to negative (Fig. 6.1). To define the constraints for each lower limb segment model only biomechanics of the same segment were used. A transition of the angular velocity from positive to negative or vice versa served as the main trigger before any of the detection constrains were applied.

Blocking times after an event detection were used to prevent multiple detections in a given time window. The constraints used were: 150 ms for the shank model, 200 ms for the leg model, and 250 ms for the extended leg model. While 150 ms helped to neglect multiple detections at heel-strike, the increased time periods were required to overcome multiple detections at toe-off.

An additional constraint was based on the absolute change of the segment angle during the stance phase. When the angular velocity became greater than -50 °/sec (during swing phase near the heel-strike event), the change in the absolute segment angle was measured. A change of 15 ° enabled the detection of the take-off event. This rule im-

Table 6.1: Rule sets applied to the different lower limb segment models with respect to heel-strike (HS) and toe-off (TO).

	Timer HS	**Timer TO**	**HS => TO => HS**	**TO if HS opposite**
Thigh	–	–	–	–
Shank	150 ms	150 ms	used	used
Leg	200 ms	200 ms	used	used
Extended Leg	250 ms	250 ms	used	–

proved detection reliability during walking, but also during standing. The threshold of the trigger was set to -50 °/sec, rather than 0 °/sec, to improve detection.

An additional constraint ensured that following a heel-strike event only toe-off can be identified and vice versa. This constraint prevented the consecutive detection of the same events. Another restriction included the opposite leg. Take-off can only happen if heel-strike of the opposite leg happened in advance. This condition ensured that there was always one leg in stance. This rule relied on the detection of double-support when both legs are on the ground. Due to the timing of the event detection in the extended leg segment model, this rule could not be applied. The extended leg segment model detected take-off before heel-strike of the opposite leg.

Table 6.1 summarizes the described rule sets. While we were able to allow only one transition from negative to positive angular velocity and one in the other direction for the shank, leg, and extended leg segment, we were not able to limit transitions with the same rule sets for the thigh segment.

6.4 Study results

6.4.1 Temporal progression of angular velocity

The grand means of the angular velocities recorded by the IMUs and the motion capture system were comparable regarding the shape and magnitude for all lower limb segments (Fig. 6.5). For both measurement methods an increase in walking speed resulted in higher peak angular velocities. We found that all lower limb segment models (thigh, shank, leg and extended leg) showed a positive angular velocity during the first part of the gait cycle and a negative angular velocity in the second part. Sign changes from positive to negative angular velocity occurred between 50% and 70% of the gait cycle. The transitions for the thigh model occur first in the gait cycle whereas the shank model marks the last transition (see Figure 6.5). Sign changes from negative to positive occurred between 80% and 10% of the gait cycle. The thigh model marked the first and last transitions in the gait cycle due to multiple zero crossings.

6.4.2 Angular velocity and gait phase detection

When comparing both measurement methods, motion capture (Fig. 6.6, left) and IMU (Fig. 6.6, right), the least amount of angular velocity sign changes per stride were found

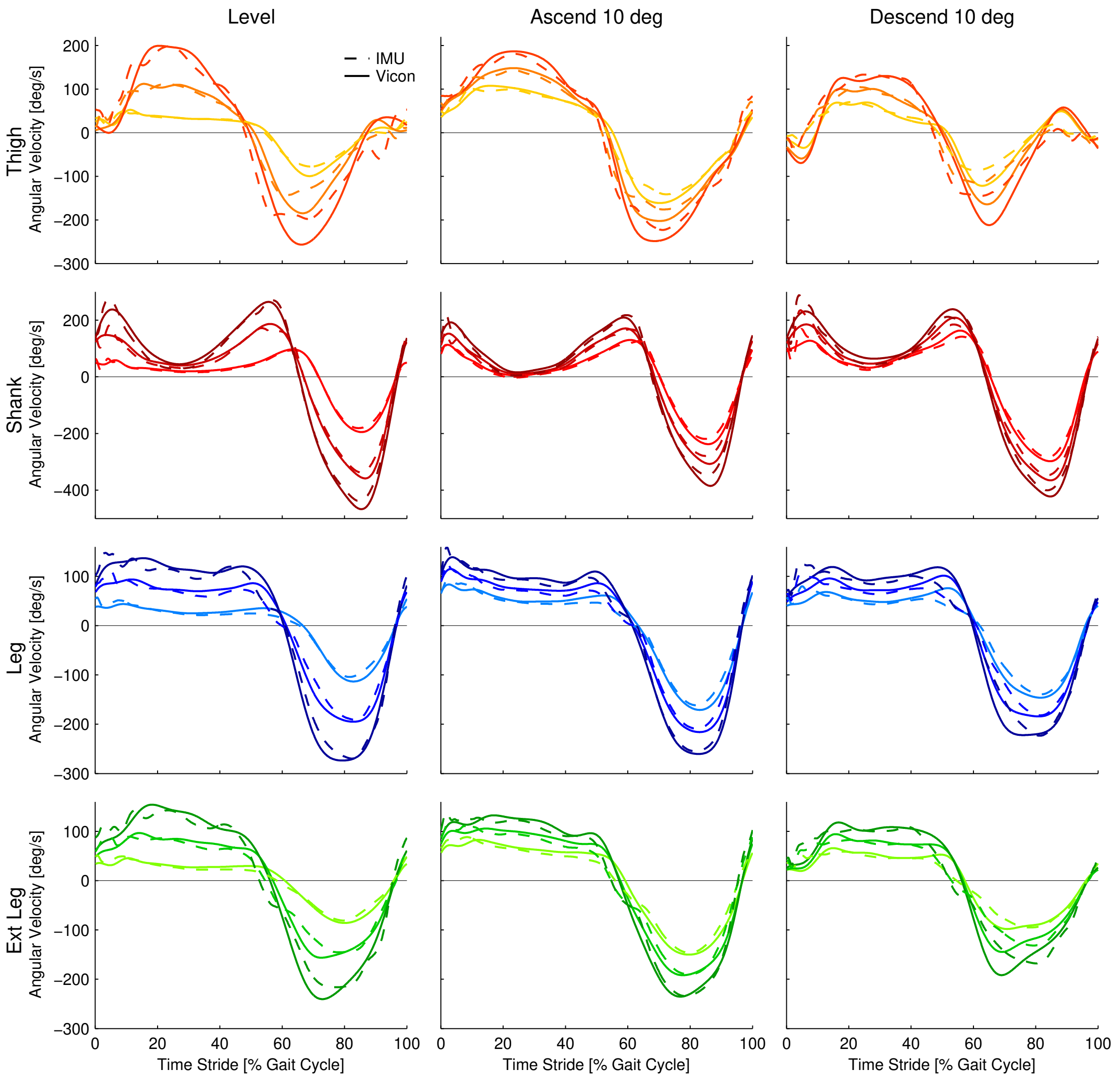


Figure 6.5: Angular velocities from motion capture (solid) and inertial measurement units (dashed) for the thigh (orange), shank (red), leg (blue) and extended leg segment model (green) in level walking, inclined walking (10°) and declined walking (10°). Darker colors indicate higher speeds (0.9, 1.3 and 1.7 m/s for slopes, 0.5, 1.3 and 2.1 m/s for level walking).

for the filtered motion capture data. All four lower limb segments had an increased amount of angular velocity sign changes if they were determined based on IMU data. Other than the grand means (Fig. 6.5), individual single stride IMU data (Fig. 6.7) could reveal signal details that resulted in the increased amount of angular velocity sign changes. While the least changes in sign were identified for the motion capture data of the leg and the extended leg, both showed oscillation around zero in several subjects and conditions at about 60% of the gait cycle when using the IMU data.

When comparing both measurement methods, motion capture (Fig. 6.6, left) and IMU (Fig. 6.6, right), the least amount of angular velocity sign changes per stride were found for the filtered motion capture data. All four lower limb segments had an increased amount of angular velocity sign changes if they were determined based on IMU data. Other than the grand means (Fig. 6.5), individual single stride IMU data (Fig. 6.7) could reveal signal details that resulted in the increased amount of angular velocity sign changes. Circles indicate oscillations in IMU based data that can cause unwanted sign changes in the angular velocity. Representative strides were selected from one subject and condition with the highest amount of zero crossings shown in Fig. 6.6. While the least changes in sign were identified for the motion capture data of the leg and the extended leg, both showed oscillation around zero in several subjects and conditions at about 60% of the gait cycle when using the IMU data.

6.4.3 Transition timing

Based on our concept shown in figure 6.1, only one sign change of angular velocity should occur at heel-strike and one at toe-off. As multiple sign changes occurred for all lower limb segments, rule sets were introduced to limit the detections to just one sign change per velocity direction and stride. The identification of the biologically relevant, single zero crossing can be achieved by comparing the IMU based data to the timing of the heel-strike and take-off obtained from force plates (Fig. 6.8). The standard deviation is indicated by the vertical line. As multiple zero crossings occur for all segments, specific detection restrictions were designed to only detect one transition for each event. Positive values indicate a zero crossing after the GRF based event detection and negative values a transition before the GRF based event detection. Due to multiple zero crossings of the thigh segment in each stride (Fig. 6.5), we were not able to limit the amount of detections with our rule sets as desired. In consequence, we were not able to compare the timing of the sign changes of thigh angular velocity with the timing based on the vertical GRF. The comparison was possible for the shank, the leg and the extended leg segment.

IMU-based zero crossings of the shank, leg, and extended leg segments occurred about 3% to 5% earlier than the heel-strike event based on GRF, while the standard deviation was lowest for the shank segment. In contrast, the timing for the toe-off deviated in both directions in all three segments. For the shank segment it occurred 4% to 7% after toe-off, for the leg from 3% before and up to 2% after toe-off, and for the extended leg from about 9% to 5% before toe-off. The shank segment showed the smallest standard

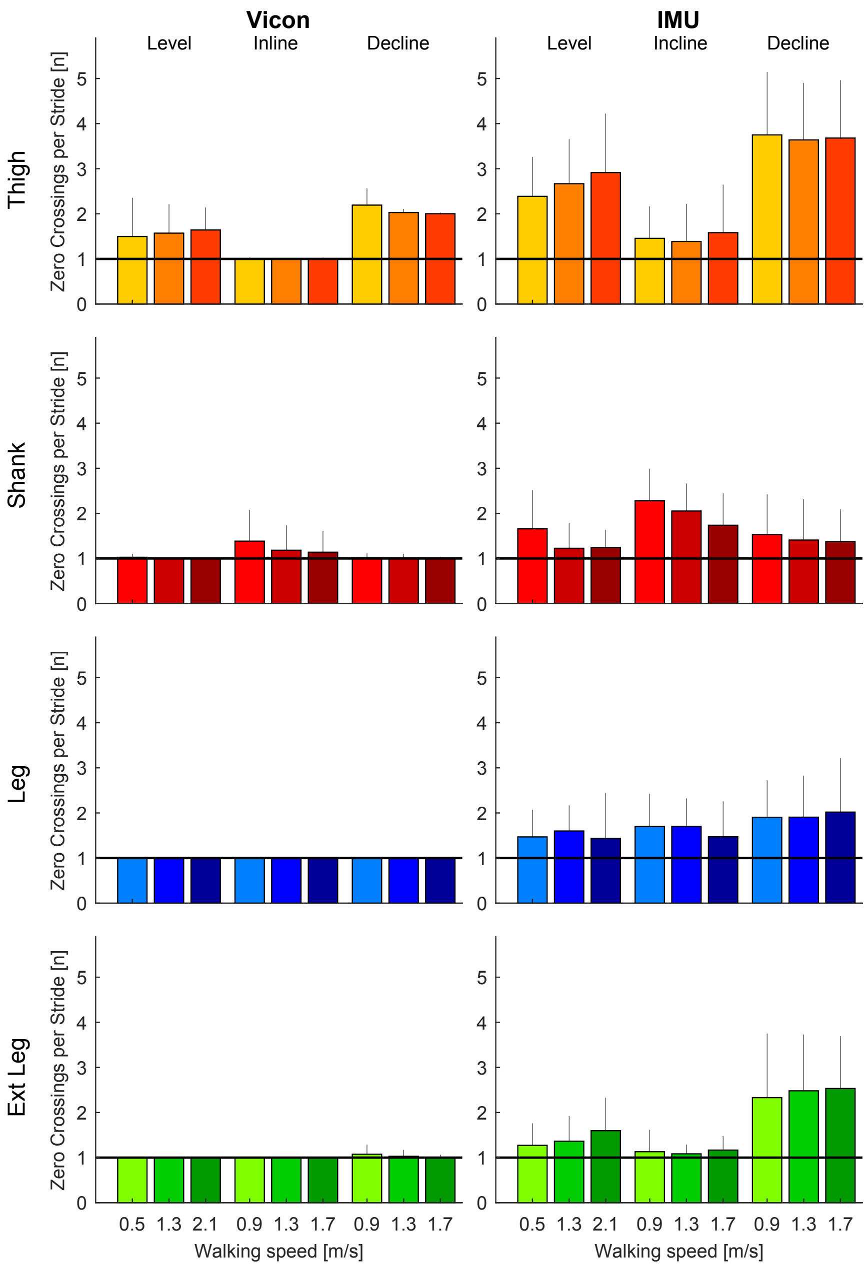


Figure 6.6: Amount of zero crossings of the angular velocity per gait cycle for the thigh (orange), shank (red), leg (blue) and extended leg segment model (green) in level walking, inclined walking (10°) and declined walking (10°). Darker colors represent higher speeds (0.9, 1.3 and 1.7 m/s for slopes, 0.5, 1.3 and 2.1 m/s for level walking). Lines show the standard deviation.

deviation for the toe-off. While an increase in walking speed had no significant effect on the timing of zero crossings at heel-strike, it had an effect regarding toe-off. A temporal shift towards later stages of the gait cycle was identified.

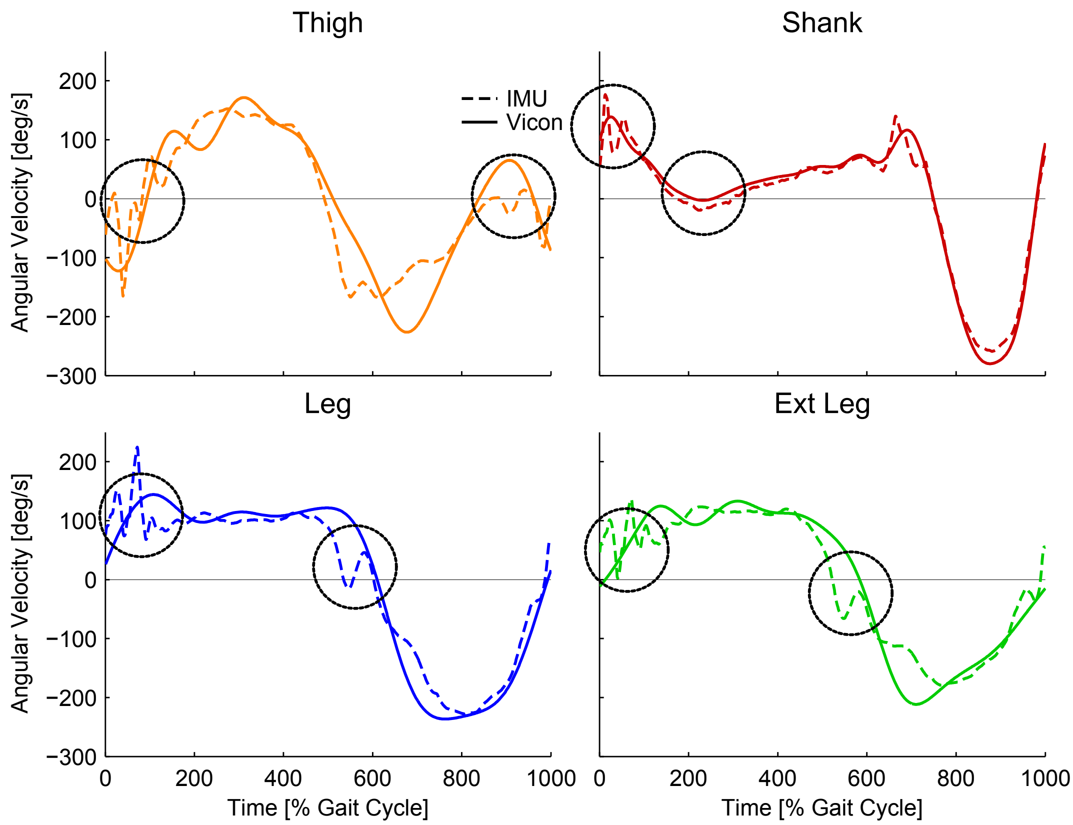


Figure 6.7: Single subject angular velocities for one stride from motion capture (solid) and inertial measurement units (dashed) for the thigh (orange), shank (red), leg (blue) and extended leg (green) segment model.

6.5 Discussion of segmentation options

We evaluated four lower limb segments models (thigh, shank, leg, and extended leg) to reliably detect gait phases (stance and swing)and the timing accuracy for gait events (heel-strike and toe-off). We based this detection approach solely on the angular velocity of biological and virtual lower limb segments during ground-level and slope walking. We further compared how these models performed when the signals were recorded a motion capture system and an array of IMU sensors. The reliability when detecting gait phases was best when using the virtual leg and the extended leg segments. The accuracy for gait events was highest for the shank segment when detecting heel-strike. For the toe-off event, improved methods are needed as either the event was consistently detected too late, or the variability in the timing was too high across models.

6.5.1 Reliability

All lower limb segments showed similar trends for the angular velocity when based on either motion capture data or IMU data. However, the filtered derivative of the angles were superior in their reliability when using the optical motion capture data. This was evident by fewer sign changes. The IMU based angular velocities generally showed more

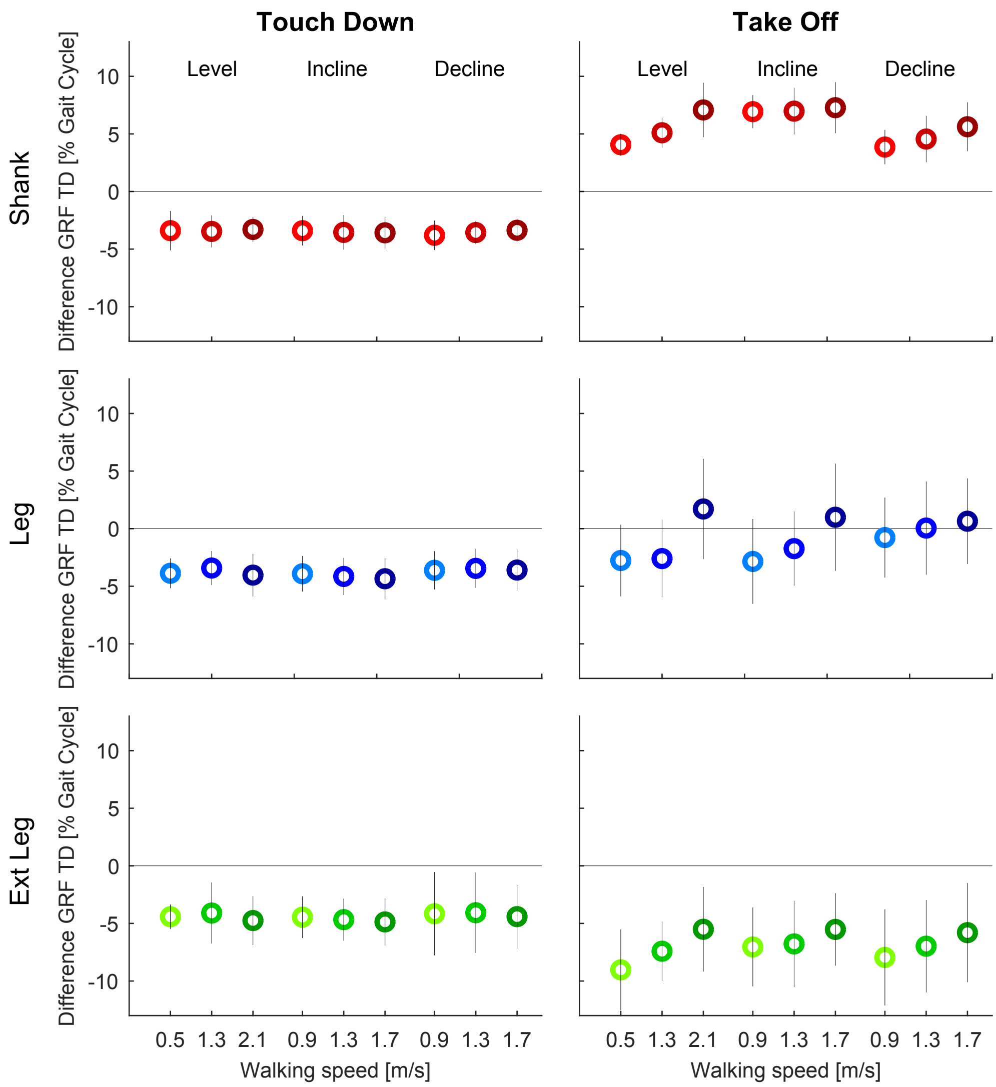


Figure 6.8: Distance of the mean zero crossings of the angular velocity based on the inertial measurement unit to the heel-strike (left) and the take-off (right) identified by ground reaction forces (GRF). Evaluated segment angular velocities from the thigh (orange), shank (red), leg (blue) and extended leg (green). Distances are evaluated for three different speeds of level walking, walking inclines and walking declines. Darker colors indicate higher speeds (0.9, 1.3 and 1.7 m/s for slopes, 0.5, 1.3 and 2.1 m/s for level walking).

oscillations in the signal, which in turn resulted in wrong detections. Independent of the origin of the data, the various lower limb segments models did not perform equally well with respect to their reliability. The best model, to reliably distinguish the gait phases based on filtered motion capture data, was the virtual leg and the extended leg segment with a preference for the leg segment. These segments showed single zero crossings close to the biological transition from stance to swing phase and vice versa. In contrast, the shank and thigh segments showed multiple zero crossings at these points, even when based on motion tracking data. The performance of the models decreases even more when using IMU data. Based on our results, the idea of using the sign of the angular velocity of a segment for stance and swing detection had to be be refused. One reason for the low reliability when using a segment is that the mechanical impact of the heel-strike event causes oscillations that affect the reliability of the algorithms (Fig. 6.7). Multiple detections near toe-off are induced by the opposing movement directions of the shank and the thigh that lead to oscillations of the angular velocity in the leg and the extended leg segments. In addition, oscillations at toe-off may be a result of the opposite limb's heel-strike event. As both, shank and thigh sensor signals were combined for the leg and the extended leg segment the signal noise of both IMUs is aggregated.

6.5.2 Accuracy

The reliability assessment showed that with a portable IMU setup, as used in our measurement, undesired angular velocity sign changes were identified. For this setup, additional constraints were needed to limit the sign changes identified to only one per direction, during each stride. These detection constraints were tuned and applied to the thigh, the shank, the virtual leg, and the extended leg. The thigh segment was excluded as we were not able to define rule sets to limit the sign changes for this segment as desired. The inability was a result of oscillations that caused multiple transitions that were not around the heel-strike and toe-off events. We conclude that the angular velocity of the thigh is unsuitable for stance and swing phase detection using our setup. However, additional constraints may improve the signal quality and render the thigh a reliable source to differentiate gait phases. When using the additional rule sets for the shank, the virtual leg, and the extended virtual leg, we found a systematic bias to early detection for the heel-strike event. Different timings were identified for toe-off. For wearable robotic system an on-time or early detection is often important to ensure a force-application at the right moment, whereas late detection could cause an unfavorable one. For example, the Myosuit [122] could use the early detection to pre-tension the system for an upcoming force application. In direct comparison, the late detection of the toe-off of the shank segment seems problematic to be used in an application. The on-time detection of the toe-off using the virtual leg and the early detection of the extended leg segment seem suitable for a controller of a wearable robot.

The variability in gait segmentation and event detection poses an additional challenge. The lowest standard deviation for detection of the heel-strike event favors the shank segment model. On the other hand, the preferred models for heel-strike detection–leg and extended leg segment–show the largest variability. This effect is larger at higher walking speeds; thus the virtual leg model could cause a late detection of the take-off events. The study identified the shank segment model as the most accurate model to detect the heel-strike event. An improvement of the methods seems required for the detection of toe-off as it was detected either too late or the variability in the timing of detection was too high.

6.5.3 Limitations of detection methods

Lower limb segment velocity, originating from motion tracking data, can be seen as an indicator that our concept can work reliably to distinguish stance and swing phase of walking. When testing the IMU data we found several critical issues. To ensure a more reliable and accurate detection, the methodological approaches could be changed or improved. The following paragraphs introduce possible options.

The signal noise of the IMU-based measurements may be strongly dependent on the placement and the interconnection to the human body. While markers were lightweight and placed on the relatively stable skin on top of bones, the IMUs were placed inside of a pocket on straps on top of the muscles and tendons. The increased inertia of the IMUs and the compliant interface to the human body may have caused an increase in signal noise. One approach to reduce noise could be to fix the IMUs to the assistive device itself.

Reliability may also be improved by using additional sensors. An IMU at the foot was successfully used in [176] to detect the transition from stance to swing. The distinct peak before the zero crossing of foot angular velocity during stance served as a feature to improve toe-off detection. It was demonstrated that the angular velocity and acceleration of the foot were more reliable than the shank angular velocity in a patient population. The peak angular velocity of the foot has to be explored in subsequent studies to investigate if it can improve detection in different walking conditions.
Alternatively, ground reaction force sensors or foot switches could be used to further improve the detection. While IMUs are required for the unique controller of assistive devices like the Myosuit [122], GRF sensors or foot switches would increase the complexity of the system with respect to donning and doffing procedures, washing, or user adaptations. For other systems, that do not require IMUs, such sensors could potentially reduce the complexity.

A reduction of the oscillations that caused multiple zero crossings could be achieved by implementing filtering techniques. Since online filtering introduces a delay, it is desirable to use the signal of a lower limb segment that depicts zero crossing before the take-off occurs. The angular velocity of the extended leg segment has exactly that property. In such case, the filter would not only prevent high frequency oscillations, but also shift the timing of the signal closer to the actual gait event measured by the GRF. In a preliminary test, a second order Butterworth filter with a cut-off frequency of 4 Hz was applied to the IMU signal of the extended leg. While standard deviations were only reduced slightly, the amount of additional detections were reduced to half.

Other improvements could include changes or additional detection rule sets and constraints. The detection rule set that showed a significant improvement in avoiding physically-irrelevant detections around take-off made use of the aphasic movement pattern of opposite limbs during walking. Take-off detections were only enabled when heel-strike was detected for the opposite leg. This ensured double support during walking and served as one of the triggers for take-off.

It has to be noted that all study participant were unimpaired. People with an impairment show different kinematic patterns that affect the detection capabilities of a chosen approach. In addition, a powered assistive device, as well as overground walking, likely affects the kinematics of the lower limbs. Thus adaptations of the approach may be required to differentiate the gait phases stance and swing.

6.6 Summary of IMU based gait segmentation

This study aimed at evaluating if it is possible to reliably detect stance and swing phase based on the sign change of the angular velocity of biological (thigh, shank) and virtual (leg, extended leg) lower limb segments during walking. In addition, the study evaluated the timing of the sign change and thus the accuracy of the detection for the transition from stance to swing (take-off) and from swing to stance (heel-strike). The study included 13 unimpaired participants and was performed at different inclinations and walking speeds to check the sensitivity of the approach to different walking conditions.

Based on optical motion tracking, we found that using the sign of the angular velocity of limb segments for stance and swing phase detection is possible. However, a reliable and accurate algorithm requires additional constraints (rule sets) based on temporal and causal relations that can be found in walking patterns. This is particularly important when using such a phase detection approach in an online application using IMU-based measurements. Rather than using the sign of the velocity on its own, we found that it is required to detect the events of the sign changes (heel-strike and toe-off) to separate stance and swing phase.

For the detection of heel-strike events–and thus the beginning of the stance phase–the shank, virtual leg, and the extended leg models were identified to all be good candidates. The most promising model used the angular velocity of the shank segment as it showed the least variability in detection timing. The thigh segment was excluded from the analysis of timing accuracy due to too many wrong detections. For the detection of toe-off events, we did not find a convincing candidate. Detection timings changed with increasing speed and variability was rather high for both candidates, the virtual leg and the extended leg.

Chapter 7

Feasibility test

The development of the Myosuit has been taken over by the ETH spin-off MyoSwiss AG after the development of the first prototype presented in chapter 5. That is why the first feasibility test has been conducted with the alpha prototype provided by MyoSwiss. Before the presentation of the feasibility test, the Myosuit alpha will be introduced. The focus will be on the differences compared to the first Myosuit version and the related set of requirements and design principles.

7.1 Myosuit alpha

The Myosuit alpha is shown in Figure 7.1. Even though the system architecture has been altered by changing the placement of the actuators to the back of the user, all the essential requirements (table 3.1) are still met. The derived principles related to the interface, provision of force and controls are also all fulfilled except for the one related to wheelchair use. The design of the Myosuit alpha does not allow to sit comfortably in a wheelchair over long times . The requirement related to wheelchair use caused the actuators to be attached distally on the shins initially. Even though the first actuators were comparatively light-weight $(1070g)$, the added mass was clearly noticeable even during slow walking. This reflects the main reason why the actuators were shifted proximally to a backpack that also contains all the electronics and the battery. The Myosuit alpha provides an upper body harness to carry the backpack but also to extends the anchoring to the upper body. The actuators on the shin have been replaced by a light-weight knee brace that is guiding the tendon over the knee and attach to the shank and thigh. The system uses the same motors and similar gear ratio. Also the passive elements at the knee and thigh remained. The knee brace provides an attachment for a strap that can be connected to the shoelaces to keep the foot in a neutral position during swing phase. The cable travel provided by the device has been extended to $45cm$ since the cable travel of the previous version was too short for some users. The device weight increased slightly to about $5kg$ compared to the $4.56kg$ of the previous version.

In chapter 6 it has been concluded that the shank velocity can be used to reliably detect heel strike. We found that this parameter can also be used in different patient populations. As soon as a heel strike is detected, the exoskeleton starts to provide forces that are based on linear scaling of the knee angle. The release of the forces is dependent on

Figure 7.1: Myosuit alpha developed by MyoSwiss AG.

a predetermined hip angle. The force magnitude and release point during cyclic movements can be adjusted by a licensed physiotherapist using the interface on the backpack. The same interface can also be used to switch between the modes for cyclic activities like walking and stair negotiation as well as sitting transfers.

To avoid reliance on tissue compression for hip and knee angle estimation, IMUs have been added to the thigh brace. Consequently, the control chart in figure 5.5 changes slightly. Figure 7.2 shows the updated control chart. The knee angle to scale the

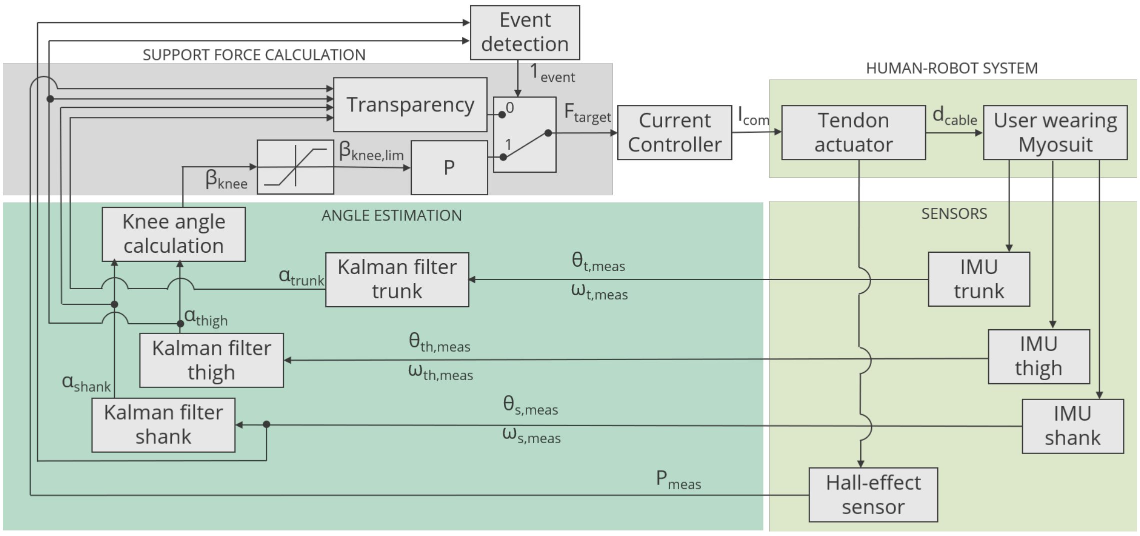


Figure 7.2: Control diagram for the Myosuit alpha.

forces can be calculated from the IMU directly and does not require the compression compensation and the feedback loop related to the encoder counts. Another change affects the closed-loop force feedback control. The Myosuit alpha does not use loadcells anymore. The main reason behind the change was driven by MyoSwiss to reduce the production cost and to increase system stability. During stance phase, the provided forces are estimated directly from the motor current. In swing phase where the exoskeleton should be transparent, MyoSwiss developed a proprietary algorithm that uses IMU data in a closed-loop system to regulate tendon length ("transparency", figure 7.2). The algorithm allows the tendon to track user movements closely and to apply forces quickly without winding an excessive length of cable before force application [177]. These modifications do not change the system behavior and the control principles demonstrated in the previous chapters.

7.2 Test protocol

The feasibility test provided a first evaluation regarding the use of the Myosuit within a clinical environment. The sessions were guided and performed by a licensed physiotherapist (PT). The PT was responsible for the organization of each therapy session. The therapeutic activities were planned and conducted according to the deficits of the participants and include strength, balance and endurance exercises. The PT received training on how to use the exoskeleton which also included the procedure of donning and doffing the device. The feasibility test aimed to investigate the practicability and the acceptance of the Myosuit.

The test was conducted at Balgrist Campus Zurich with approval by the Research Ethics Committee of ETH Zurich. Patient recruitment was lead by the PT and informed consent was given by all participants before the start of the test. Participants were selected according to the following inclusion criteria:

- between 18 and 70 years old
- BMI less than 30
- being able to move the lower limbs against gravity independently
- being able to perform sitting transfers and walk 10m independently (conventional mobility aids allowed)
- Montreal Cognitive Assessment test (MoCA) more than 24 points

As part of the screening process, potential participants had to fill out the MoCA test [178]. The test is able to detect mild cognitive impairments and Alzheimer's disease. It is a requirement to be cognitively unimpaired to use the Myosuit alpha. The test protocol is depicted in figure 7.3. The whole test was conducted over six weeks with one session of 45 min each week. Each participant had one familiarization session with the PT to get used to the device and to find the optimal use parameters such as sizing, force magnitude and timing of swing phase detection.

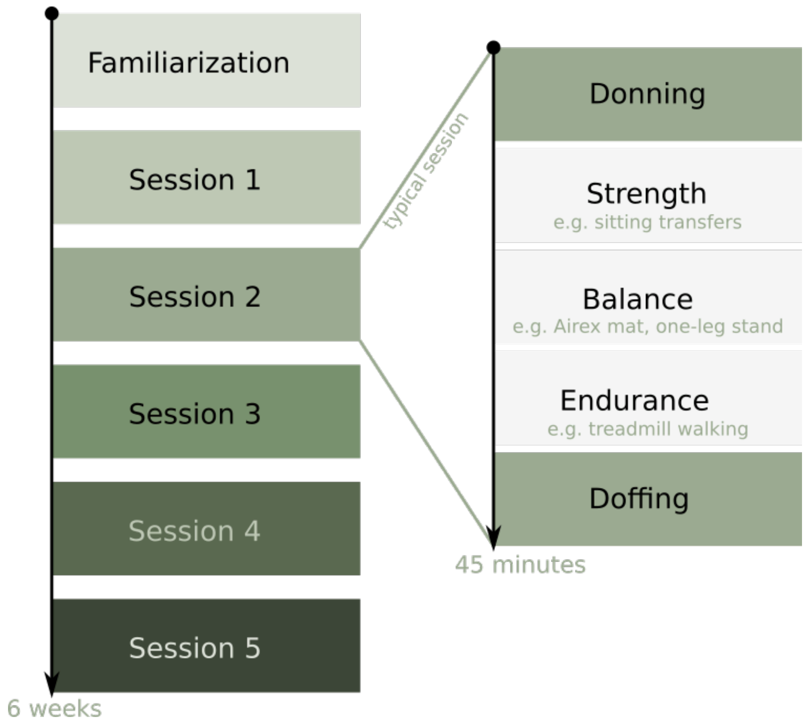


Figure 7.3: Protocol of the feasibility test.

All subsequent sessions followed a similar pattern. A session always started with donning the suit and ended with doffing it. Depending on the therapy goals of the participants, the PT conducted specific exercises that could be assisted by the Myosuit.

Independent of the chosen exercises, the therapy always comprised three core elements related to strength, balance and endurance. All exercises were performed in a activity related manner. Strength training included moving the participant's body against or with gravity facilitating concentric and eccentric muscle contractions without challenging the participants cardiovacularly. Strength training could consist, for example, of multiple repetitions of sitting transfers with sufficient breaks between sets. Endurance training consisted of prolonged walking resulting in self-reported, elevated activity of the cardiovascular system over an extended time period. The time period was determined by the PT. Balance training targeted isometric muscle contractions while the participants had to focus on different proprioceptive tasks introduced by the PT. Balance exercises were usually done while standing on one or two legs, with or without the use of other therapeutic equipment. The PT used and adjusted the assistive forces of the Myosuit, so that patients perceived the support during the exercises.

To assess the practicability of the Myosuit, the subsequent measurements were taken:

- time for donning and doffing
- System Usability Scale (SUS)
- adverse effects

Since time is very limited in everyday physiotherapy sessions, it is important that the Myosuit can be donned and doffed quickly. By talking to several physiotherapist before conducting the feasibility test, we found that 5 min for both processes seemed to be accepted. Therefore, the practicability partially depended on whether the PT is able to don and doff the Myosuit within this time. The time was measured at every session.
The SUS is used to measure the usability of a device [179]. The usability questionnaire uses a five-point Likert scale from "strongly agree" to "strongly disagree" and has ten items in total. The result of the SUS is a weighted score between 0 and 100 points. The minimum score that the Myosuit should achieve is 60 points. A score of less than 50 indicates that a device is not usable enough. The questionnaire was provided to the PT and the patient in the session one, three and five.

In the framework of this feasibility test, adverse effects comprised any untoward occurrences related to user safety or device usage. Adverse effects had to be recorded when they occur. They were divided into three groups: light, moderate and severe. The PT was responsible to assess the adverse effects accordingly. Light adverse effects relate mainly to technical problems that can be solved within the same session. These effects usually do not affect the performance of the device. Moderate adverse effects comprise technical issues that can not be solved by the PT and affect the performance of the device but not patient safety. Severe adverse effects relate to technical issues that cause harm on the user, and thus, directly relate to user safety. To ensure that the Myosuit is safe to be used during therapy sessions, no severe adverse effects should be observed during the course of the feasibility test.

The assessment of the acceptance of the Myosuit is dependent on the following measurements and observations:

- Würzburger Erschöpfungs-Inventar (WEIMuS)
- skin incidents
- Numeric Rating Scale (NRS)

WEIMuS provides a measure in form of a questionnaire to quantify fatigue [180]. It includes 17 questions that are scored from zero to four. The maximum number of points is 68 points. Higher scores indicate a high level of fatigue, whereas low scores indicate low levels. The tool measures cognitive and physical fatigue. WEIMuS were used to detect if there is an increase of fatigue score over subsequent sessions that might indicate too intensive therapy sessions. The tool was used after each session.

Skin incidents refer to skin integrity and sensitive pressure points. Any skin irritation or damage were classified by the PT as light, moderate or severe. Light skin incidents are marks that disappear within 15 min and cause no discomfort or pain to the user. Moderate skin incidents are marks that cause pain or discomfort and disappear within 60 min. Skin integrity is maintained for both levels. In contrast, severe incidents damage the skin and take considerable time to heal and cause discomfort or pain for multiple days. No severe skin incidents should be observed during the course of the feasibility study.

Pain assessments can be done by using the NRS. Each participant was asked according to the NRS before, during and after the therapy session. The score ranges from one to four and an improvement of 0.5 points is deemed to be statistically relevant. Patients should no report any longstanding pain related to the use of the Myosuit.

7.3 Test results

Nine patients have been screened according to the inclusion criteria. Four of them had to be excluded.

	Mean	**SD**	**Median**	**Range**
Age (years)	51.4	6.24	52	42-58
Height (cm)	180.4	10.17	183	162-190
Weight (kg)	80.6	8.54	75	73-92
BMI	24.88	2.71	25.8	21.6-27.8

Table 7.1: Participant demographics.

One person lost interest in participating in the test. Two patients were not able to stand up by themselves and the fourth person had to be excluded due to recent and unrelated femur trauma. Hence, the feasibility testing was conducted with 5 patients. All participants were male. They scored more than 24 points in the MoCA assessment. Table 7.1 provides more information about the demographics and table 7.2 about the clinical characteristics of the participants. The maximum, self-reported walking distance was included as an indirect measurement of the patients ability to walk. Lower distances are associated with reduced walking ability and higher level of impairment.

Participant	**P1**	**P2**	**P3**	**P4**	**P5**
Pathology	Bethlem Myopathy	MS	MS	iSCI C5	Stroke
Walking distance (m)	3000	5000	20	2000	1000
Walking aid		walking sticks	rollator	walking sticks	

Table 7.2: Participant pathology, self-reported walking distance and mobility aid needed to cover the distance.

All five participants finished the test protocol and completed all sessions. None of the patients experienced serious adverse events. Figure 7.4 shows the results per session for donning and doffing time, SUS and WEIMuS.

7.3.1 Practicability

Donning and doffing

Donning times did not meet the goal of 5 min. The average donning time, however, decreased over subsequent sessions. In session one, the average donning time was 10.37 min across all participants. The average decreased by 3.41 min to 6.56 min in session five. The shortest donning time in session five was 5.35 min and the longest donning time was 8.35 min. The longest donning time over all sessions was 15.14 min for participant one at the first session. This patient also showed the greatest reduction in donning time of 9.19 min. Not all patients reached their best donning times in the last session.

All doffing times are under 5 min at session five. In session five, doffing times ranged between 2.08 min and 2.43 min. The shortest doffing was 1.47 min for participant five in session four and the longest time is 5.01 min for participant two at session one. Only one participant exceeded the doffing time of 5 min in one session.

System usability scale

The SUS score given by the participants showed a range from 60 to 100 points in session five. All participants except for P2 showed a constant increase. The highest increase over all the sessions was 17.5 points for participant one and two. The lowest increase was five points for participant five. Only one evaluation resulted in a score of less than 60 points. The evaluation of the SUS for participant two at session two resulted in a score of 50 points. Participant two reported dynamic balance issues due to the Myosuit assistance and related noise of the motors which caused a deviation from his usual walk-

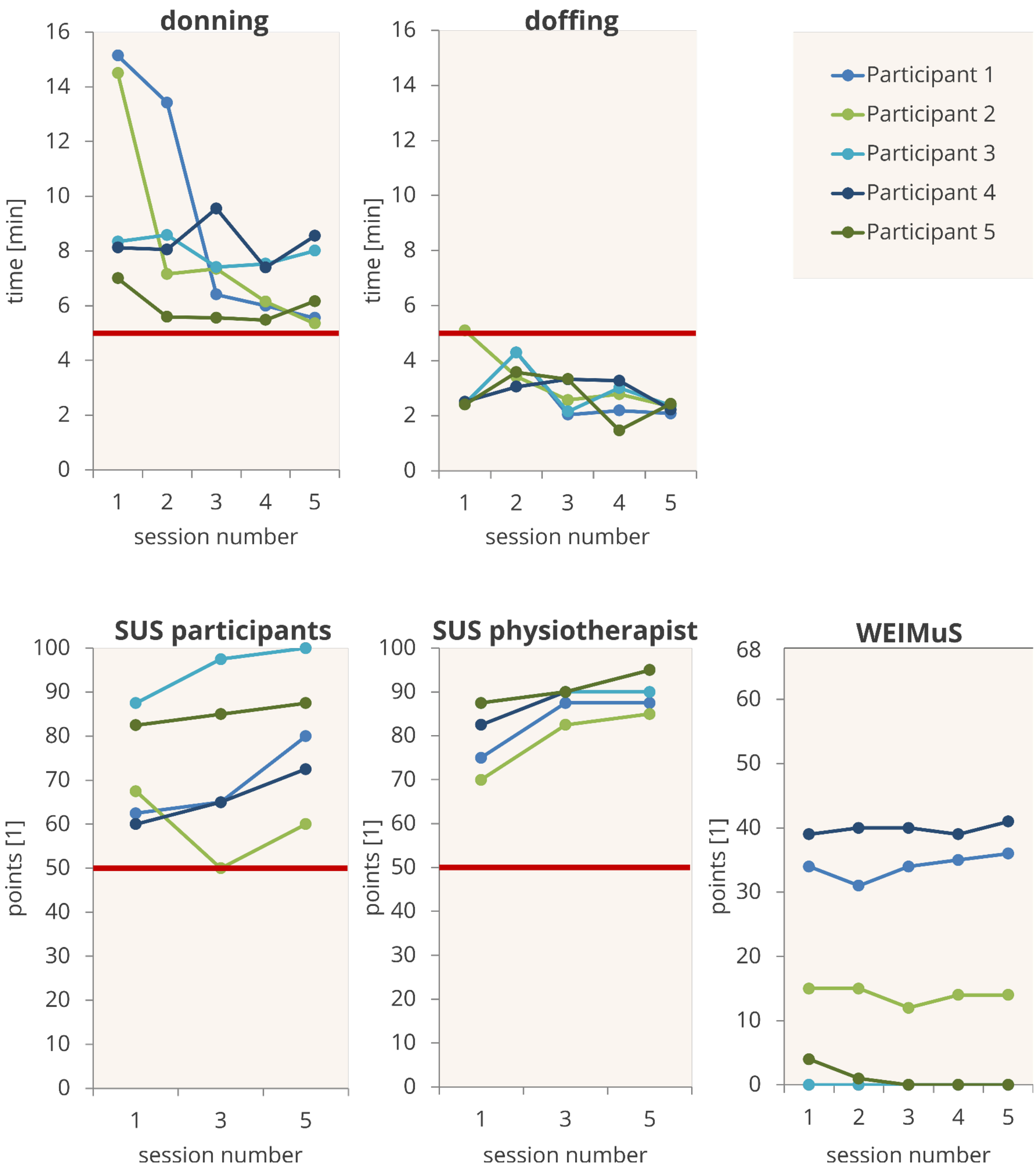


Figure 7.4: Results of the feasibility test (donning and doffing time, SUS, WEIMuS).

ing rhythm. This was reflected in a lower SUS score.

The SUS score given by the PT ranged from 85-95 points across all participants in session five. Scores for all patients increased over subsequent sessions. The highest increase was 15 points for participant two and the lowest 2.5 points for participant three.

Adverse effects

No moderate or severe effects were reported in the feasibility test. However, there have been a number of light adverse effects described. Table 7.3 describes these effects and also indicates who reported them.

Light adverse effect	Reported by
Tech. issues (resolved by PT according to instructions)	PT (8 times)
Usability error related to force magnitude setting	PT (6 times)
Usability error related to swing phase calibration	PT (6 times)
Sound caused disturbing sensation	P1,P2,P4
Wearing the Myosuit caused strange sensation	P2
Usability error (i.e. missed steps) during donning procedure	PT
Knee braces touch medially	P3,P5
Early heel strike detection	P4

Table 7.3: Light adverse effects reported.

7.3.2 Acceptance

Würzburger Erschöpfungs-Inventar

The WEIMus scores were generally stable for all participants. Considerable increases between the subsequent sessions would have indicated increased fatigue and too intensive training. No considerable changes were detected.

Skin incidents

All participant reported light skin incidents in form of pressure marks that disappeared within 15 min. However, two moderate events were reported. Participant one exhibited pressure marks from the Velcro straps of the knee brace after session one. The mark and the red skin color disappeared after 40 min. Participant four showed a pressure mark from the padding of the knee brace that was folded unintentionally after session four. The pressure mark and the red skin color disappeared after 60 min.

Numeric rating scale

There were no reports of pain before, during or after a therapy session with the Myosuit.

7.3.3 Unstructured feedback

The PT and patients provided feedback after the feasibility test. It has been observed that three participants felt that the training with the Myosuit improved their endurance. These patients reported that they were able to walk longer distances than before. Improvements with respect to balance have been noticed in the other two participants. They described improved balance during sitting transfers and that they could stand longer without help. Their regular physiotherapists also commented that they perceived positive changes related to endurance, spasticity and range of movement at the hip and knee joint. The PT that conducted the tests noticed that patients showed improvements in their walking pattern after a session with the Myosuit. There has also been a noticeable boost in motivation in all participants. All participants reported that they would like to continue to train with the Myosuit after the study. Even though the PT reported that the Myosuit is very useful during physiotherpy sessions, the biggest downside of the Myosuit is the complexity of the donning process and the corresponding time according to the PT.

7.3.4 Summary of results

The feasibility test showed that the Myosuit can be used in physiotherapy sessions. The Myosuit supported strength, balance and endurance exercises in a diverse patient population. No serious adverse events have been reported and the usability was rated high on average by the participants and the PT. However, it has also been shown that the design of the Myosuit alpha is too complex to allow short donning times. The long donning time seems to be a hurdle to integrate the Myosuit in physiotherapy sessions outside of the feasibility study.

A discussion of the results can be found in the next chapter. The results will be put into context of the previous chapters to draw conclusions related to the support of pathological movements and the implications for devices that aim at improving these movements.

Chapter 8

General Discussion and conclusion

8.1 Discussion

There is an ever increasing number of exoskeletons that is intended to help patients with movements [23, 181]. Many researchers design these devices to guide patients through predefined movement sequences, and thus, can completely drive the kinematics of the user. These substituting exoskeletons aim to provide normal function that is primarily driven by the exoskeleton and not the user. This approach can be advantageous as it enables processes related to neuroplasticity [89]. However, they mainly address the needs of individuals that are fully paralysed or have insufficient muscle strength to move. By replacing the body function with respect to movement, these devices are required to provide high forces and a corresponding structure that can endure induced loads. This is the main reason why substituting exoskeletons are complex, tend to have high system weights, and therefore, are expensive. Nonetheless, they are useful tools for paralysed patients.

Most patients and elderly are still able to move with increased effort due to pathological movement patterns. Since these people have remaining functionality, substituting exoskeletons do not seem to be the right choice. If patients have sufficient motor function to provide pathological movement patterns, exoskeletons for partial support might be able to address their deficits directly and increase their movement economy whilst also supporting motor relearning. These exoskeletons administer only supportive forces that target certain deficits, but cannot drive the entire movement pattern. By providing lower forces and targeting deviations from normal movement at specific body locations, these devices tend to be less complex, more light-weight and less expensive than substituting exoskeletons. There are few exoskeletons for partial support that have been clinically tested in patient populations (chapter 3.2). The devices are very diverse as they target different joints and deficits. These exoskeletons are designed to increase step length [108], restore walking dynamics [115] or to provide stability [112]. The underlying reasoning for the choice of support and related forces as well as derived requirements are disparate.

To provide a basis for future and current developments, this thesis identified essential requirements for exoskeletons that provide partial support, independent of their technical implementation (chapter 3.4). To derive the essential requirements, I have identified the main mechanisms of efficient movements (chapter 2.3) and the common design principles of devices that broke the metabolic cost barrier (chapter 2.4 and 2.5). Since these characteristics do not necessarily apply to patients, mechanisms of pathological movements (chapter 3.1) as well as the common design principles of movement aids have been compared and identified (chapter 3.2). Due the disparity of support strategies of exoskeletons for partial support, two frameworks have been discussed and proposed to identify the most beneficial joints to support (chapter 3.3). The essential requirements were discussed in chapter 4 and further design principles were derived and implemented in the Myosuit (chapter 5). The Myosuit showed a reduction in activity of the supported muscles during sit-to-stand transitions of an unimpaired user. This did not serve as a full validation as it did not consider cyclic movements such as walking, nor did it involve patients. To implement different support strategies during swing and stance phase, we first had to develop a reliable segmentation method (chapter 6). This eventually enabled us to validate the essential requirements by conducting a feasibility test with five patients and diverse pathologies (chapter 7).

Designing exoskeletons to partially support pathological movements is a challenging task. Even if patients have the same pathology, it does not determine their movement pattern. Although there are common compensatory movements depending on deficits, the manifestation related to magnitude of the deviations and the combination of compensations are different for every patient. Nonetheless, I hypothesized that there are essential requirements that enable the design and development of exoskeletons for partial support that can be used with various pathologies.

The derived set of requirements has been implemented during the development of the Myosuit. Requirements

- RQ1 (light-weight design ($< 6kg$)),
- RQ2 (minimal distal masses),
- RQ3 (adjustability to different body shapes),
- RQ4 (consideration of tissue deformations),
- RQ7 (avoidance of sensitive pressure points), and
- RQ13 (partial support in the range of $4Nm - 40Nm$)

were verified or validated with the design of the first prototype or the feasibilty test. These requirements are also implemented in designs of current exoskeletons for partial support [115, 112, 109] and are in agreement with other research groups. The weight for both Myosuit prototypes is less than $6kg$ and the design reduces distal masses by relying on cable actuators and passive elements. Depending on the body composition and configuration, the Myosuit can deliver a maximum external moment between $20Nm$ and $40Nm$ at the knee and hip joint. The adjustability to different body shapes was demonstrated in the feasibility test as the Myosuit has been used with five patients of various heights and body weights. The feasibility test also showed that the Myosuit did not cause any pain. Only two cases of moderate skin pressure marks were reported due to usability errors. This indicates that the Myosuit design avoids sensitive pressure points. Tissue deformations have also been considered in the design of the cable actuator speed. Furthermore, tissue compression has even been used to estimate the knee angle in the first version of the Myosuit.

Other requirements that refer to the provision of force, i.e.

- RQ5 (Optimization of inter-segmental force transmission),
- RQ8 (Support of knee extension during stance phase),
- RQ9 (Support for flexor muscles of the hip, knee and ankle joint during swing phase), and
- RQ10 (Targeting users that are household or limited community walkers $< 0.8m/s$),

could only be validated to a limited extent. Even though the Myosuit has been tested at different speeds, the feasibility study did not control for walking speed of the participants. The feasibility study showed that the Myosuit is safe to be used and is able to provide support during physiotherapy training targeting different pathologies. The forces were not perceived as disturbing and the participants and the PT rated the Myosuit high on the SUS scale on average. However, this does not indicate that the bi-articular support of hip and knee joint and the antagonistic passive support of flexor muscles are advantageous compared to other methods. Other devices have successfully improved walking function in stroke patients by supporting the ankle or hip joint [117, 116]. This has been mainly achieved in higher functioning patients that are classified as community walkers [98]. Awad et al. reason that propulsion is a major contributor to walking-related disability [182]. Nonetheless, it has also been shown that ankle propulsion becomes less efficient with increasing knee angle [56] and that there is a strong dependency on knee extensor strength and walking speed, and therefore, functional capacity too. It seems that exoskeletons that aim to improve walking dynamics are automatically targeting the mechanics of efficient walking (figure 2.7).

Patients that struggle with weight bearing and stability, however, might not benefit from such an interventions. The deficit based approach (chapter 3.3.2) identified knee extensor support as the most promising assistance method during stance phase. This approach relies on visible deviations from normal walking. Consequently, it could be that this approach primarily considers patients that are less functioning and slower due to pronounced compensatory movements and that weight-bearing is more important to support. We have already shown in other studies that the Myosuit can increase the walking speed in patients [183, 184]. This indicates that the chosen support method is beneficial and can improve walking function. Additionally, it is important that the added value of a device is not only measured in speed improvements. Insurance companies usually expect effects that increase patient's customary level of walking ability at home and in the community. Consequently, it seems imperative that designs target slow walkers to achieve meaningful improvements in function.

The support of swinging the leg forward, specifically in parallel with the hip flexors, is consistent with the strategies successfully implemented in other devices, too [108, 109, 110]. The support of knee flexion has not been investigated so far. Swinging the leg forward constitutes a fundamental requirement for bi-pedal gait and also relates to step length which is known to be a determining factor for economical gait [3]. Compared to other knee exoskeletons, the cable actuated knee brace did not negatively affect the swing phase.

It has been shown that the identified support strategy can potentially help with the fundamental requirements of walking. The knee extensor moment plays a key role in stance phase and a flexor moment at the ankle, knee and hip during swing phase. Even though the deficit approach seemed to have resulted in requirements related to the provision of force that agrees with other research, the method has weaknesses too. Any combination of deficits is possible to be observed in a patient. This could also only include deficits that are not targeted by the main mitigations included in the essential requirements and design principles. The approach relies on the identified deviations and underlying causes that are based on data presented in the literature and might not represent a complete set of deviations and deficits. Another shortcoming is related to not considering other body planes and only deviations and deficits in the sagittal plane. The deficits in other planes could potentially also limit walking or movement function significantly. The method does not consider the severity of a given deficit either.

However, there is evidence that walking speed is linked to knee flexion and extension strength [98] which supports the findings of the presented approach. It is likely that if a deficit is more severe, walking will take considerably more effort. Patients compensate this additional effort by walking slower. It could be the case that slow patients (household and limited community walkers) benefit from support methods that help to fulfill the fundamental requirements of walking. Whereas patients classified as community walkers rather need support that can help to restore the mechanisms of efficient gait introduced in chapter 2.3.

Future studies should investigate different support methods depending on patient speeds to better understand the differentiating elements in providing exoskeletal support. The influence of the bi-articular design was only investigated on one unimpaired subject during sitting transfers. Winter and Hof described a close interrelation of the hip and knee moment during stance phase [129, 130]. Another study should investigate the effects of the bi-articular design in normal and pathological gait with respect to that interrelation.

The requirements regarding the controls of the device, i.e.

- RQ6 (Intuitive control to facilitate motor learning), and
- RQ12 (Suitable to be used as therapy and mobility device)

could also only be partially validated. The device has been successfully used within therapy sessions with a licensed physiotherapist conducting several different exercises. These exercises were activity related which also indicate that the Myosuit could be used outside of a clinical environment. The Myosuit can potentially be used in the continuum of care bridging clinical, outpatient and home rehabilitation. Patients were able to understand the active and passive support and were able to learn how to use the assistance during their movements. This is indicated by the increasing SUS scores in subsequent sessions for four out of five participants in the feasibility study. Similar results can be seen in the SUS score of the PT. However, the adjustment of the release point to allow the passive elements to assist the swinging leg (swing phase calibration) led to light adverse effects. Similar effects occurred for the setting of the force magnitude. The PT needed multiple sessions to adjust these settings optimally. Nonetheless, the physiotherapist was able to detect the errors and correct them accordingly. Another light adverse effect was associated to heel strike detection. One participant used a pass-retract movement initiated at the hip to compensate for weak hip extensors and swing the shank forward. This caused a consistent early detection of stance phase. Since the detection helped the participant to extend the leg, it was not perceived as disturbing. The feasibility test does not allow to make a statement regarding the efficacy of the posture based controller or alternative control modes. Further investigations are needed to evaluate different control approaches in patients depending on specific deviations and deficits.

The last remaining requirement was indirectly validated during the feasibility test. RQ11 (Structural design allows compensation movements) influenced the Myosuit design to not cause any restriction when worn. Patients were able to conduct balance training in the Myosuit which required also lateral movements. The participants did not report any difficulties in following the PT's instructions performing these movements. We also implemented a transparency mode that becomes active during swing phase to allow the passive elements to support a flexor pattern. This mode has been used during the familiarization session to allow patients to walk with the Myosuit, but without the application of forces. The patients reported no adverse effects nor did they perceive any restrictions. We have demonstrated in another study that the transparency mode does not affect walking kinematics [185]. However, this does not indicate that familiarization can be accelerated with such an approach. It will require a more thorough investigation to answer that hypothesis.

8.2 Conclusion and outlook

The essential requirements can serve as a first indication when designing or assessing exoskeletons for partial support. They provide a framework that includes the corresponding reasoning behind each requirement. This also includes tracebility to related research. The Myosuit is the first implementation resulting from these requirements. Even though the Myosuit has undergone preliminary testing, the related requirements are not conclusive as some points require further investigation. The Myosuit is not only a medical device but also serves as a research platform to further investigate the aforementioned aspects.

The feasibility study revealed the main weakness of the Myosuit alpha which were related to the complexity and duration of the donning process. The Myosuit alpha is a modular system that required multiple steps to be adjusted to a user. Furthermore, it was reported that the knee braces collided medially due to the design of the structure of the braces. This is resolved in the CE marked version of the Myosuit depicted in figure 8.1.

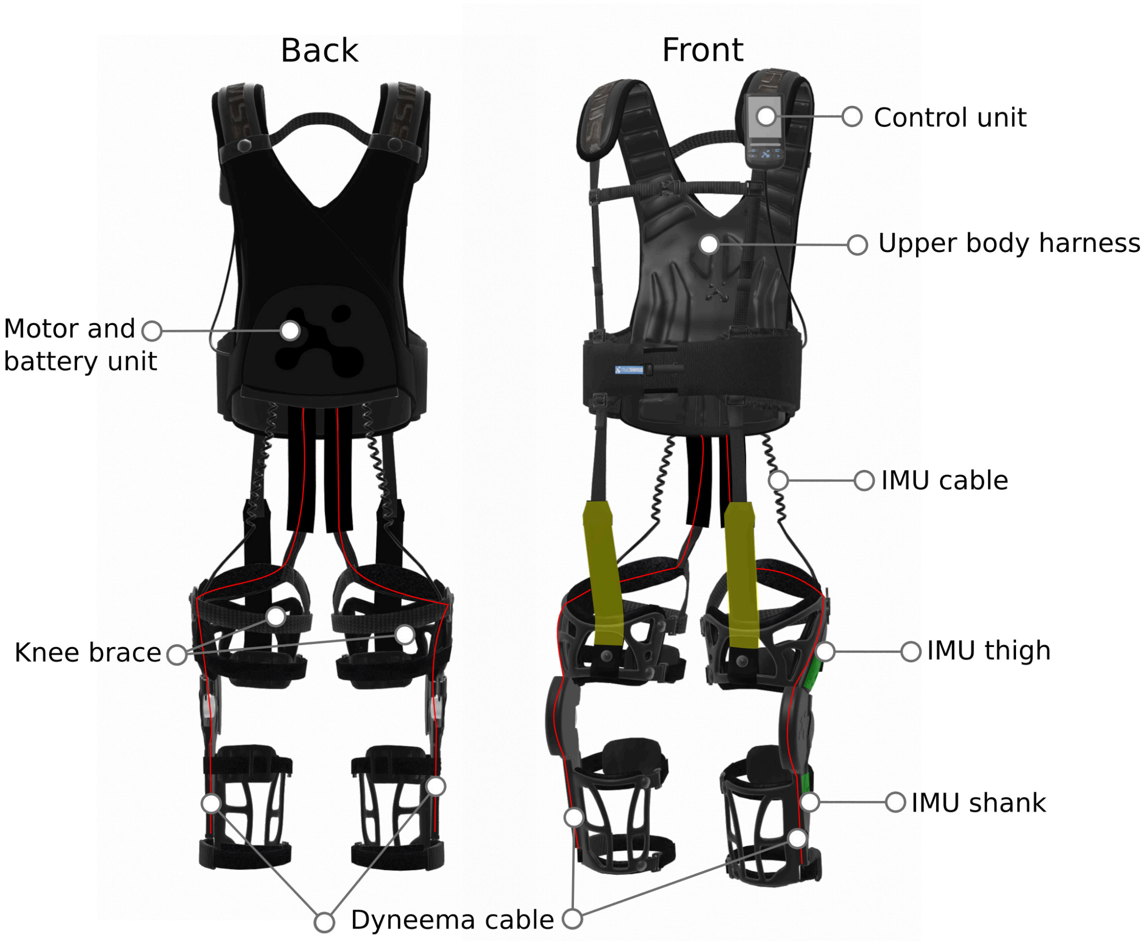


Figure 8.1: Commercial version of the Myosuit developed by MyoSwiss AG.

The main change addresses the modularity of the system. The Myosuit has no components that need to be attached or detached by the user, everything is always connected. Internal testing showed that we are able to achieve donning times between two and five minutes. This enables the Myosuit to be integrated into physiotherapy sessions or in the daily life of patients. The knee braces have also been redesigned to only use one lateral hinge joint while still guiding the cable. The design avoids the collision of the braces that has been reported in the feasibility test. It also implements a patented counter-torsion mechanism that prevents the knee brace from rotating outwards due to the pulling force of the cable [186]. Minor improvements related to system noise and the user interface of the system have also been made. Due to safety concerns, the knee ligaments to support knee flexion have been removed in the commercial version of the Myosuit. They could facilitate the collapse of the knee joint, i.e. sudden knee flexion, during stance phase when the tendon is not active. Except for the support of the knee flexors, all essential requirements are still met.

The Myosuit is a medical device that is being used during rehabilitation and as mobility aid in the European market. By following the presented essential requirements, this exoskeleton is lightweight and affordable and targets deficits that are related to weight-bearing, stability and swinging the leg forward. The field of exoskeletons lacks structure which is reflected in the high number of devices without sufficient clinical data. The provided structure and requirements could serve as a basis to design technologies that can be evaluated in a structured way. The requirements and related findings could be even extended to develop more targeted interventions or to assess current interventions based on a common framework.

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