Biomimetic Stiffness for Transfemoral Prostheses

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SERGE MANUEL PFEIFER

MSc ETH in Electrical Engineering and Information Technology, ETH Zurich

born on 16.11.1983
citizen of Zurich

accepted on the recommendation of

Prof. Dr.-Ing. Robert Riener, examiner
Prof. Dr.-Ing. Heike Vallery, co-examiner
Prof. Dr. Eric J. Perreault, co-examiner

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Abstract

The loss of a lower limb is a life-altering event, giving rise to both physical and psychological challenges. The main causes of amputation in developed nations are vascular diseases, often with a diabetes comorbidity. Trauma or tumors are other reasons that can lead to the amputation of a lower limb. The use of prostheses can restore mobility and participation of the affected people in daily life to some extent and can increase quality of life in general. However, most amputees remain severely limited in mobility compared to able-bodied people. Therefore, it is necessary to find ways to improve prosthetic functionality, which could include improving actuation technology and control strategies. A promising approach to tackling this challenge is to mimic physiological function. During physiological locomotion, humans modulate their knee stiffness continuously and subconsciously according to the demands of the activity and terrain. Given modern actuator technology, powered transfemoral prostheses could, in principle, provide a similar function. However, quantitative information of how humans modulate knee stiffness during gait is lacking.

The goals of this thesis are to quantify such physiological stiffness modulation, to determine requirements for the design of a biomimetic transfemoral prosthesis, to realize such a device, and finally to experimentally assess the feasibility of using such a device as a prosthesis.

Quantifying physiological joint stiffness commonly involves perturbing the respective joint in a controlled manner through an actuated orthosis and measuring the resulting displacement and moment at the joint. This is challenging during natural gait without impeding it substantially. In this thesis, a model-based approach is developed to estimate joint stiffness in a less disruptive manner using electromyography (EMG) combined with kinetic and kinematic measurements. The approach includes a musculoskeletal model, and it estimates muscles forces based on measurements of knee moment, kinematic configuration and EMG; a model of muscle and tendon stiffness depending on muscle force allows estimation of joint stiffness. The approach is evaluated in isometric conditions, in which experimental measurements are more practical than during gait. The validation shows that model-based estimates of knee joint stiffness coincide well with experimental data obtained using conventional perturbation techniques. The model-based approach is also applied to estimate stiffness during gait.
Based on this approach, stiffness requirements for knee prostheses are derived, and a suitable actuation concept is proposed. The idea is to express stiffness and moment requirements as functions of joint angle and then to combine a series elastic actuator (SEA) with an optimized nonlinear transmission and parallel springs to reproduce the profiles. This ANGle-dependent ELAstic Actuator (ANGELAA) is realized in a prototypal knee prosthesis that uses rubber cords as series elastic elements, effectively making it a series visco-elastic actuator (SVA). The incorporated mechatronic principles may pave the way for cheaper and lighter actuators in artificial legs and in other applications where stiffness requirements depend on kinematic configuration.

The developed prototype is tested during treadmill walking with an amputee subject. Two different control strategies employing different stiffness profiles are tested, with one of the control strategies being based on the stiffness estimated using the model-based approach. In the familiarization phase, this stiffness was required to be lowered in order to enable comfortable ambulation for the amputee. Following this adjustment, the amputee was able to walk comfortably with near-natural gait kinematics. The second control strategy was developed using manual tuning of parameters, rather than using physiological stiffness estimates; it also enabled the amputee to walk with kinematics similar to physiological gait, even though stiffness profiles deviated substantially from the profiles used in the first control strategy. The findings presented in this thesis may ultimately help to improve the quality of life of amputees.
Kurzfassung


Die Ziele dieser Arbeit sind solch physiologische Steifigkeitsvariation zu quantifizieren, Anforderungen für die Konstruktion von biomimetischen Oberschenkelprothesen herzuleiten, ein solches Gerät zu realisieren, und schließlich die Verwendbarkeit eines solchen Geräts als Prothese experimentell zu untersuchen.

Lenkssteifigkeit gut mit den experimentellen Daten übereinstimmen, welche mit konventionellen Perturbations-Techniken erhoben wurden. Der modellbasierte Ansatz wird auch auf SteifigkeitsSchätzungen während des Ganges angewendet.


Introduction

1.1 Motivation

The loss of a lower limb presents severe physical and psychological challenges for the affected person. The number of amputees can only be estimated, and is hard to identify globally [1]. The prevalence of major lower-limb loss (above knee or below knee) in the United States was estimated to be around 623’000 in 2005 [2]. For European countries, no reliable numbers for prevalence are available. Regarding the incidence of amputations in Germany, the number of major lower-limb amputations due to vascular diseases was 15’193 in 2010 [3]. With the aging population in the US, amputation rates are predicted to rise substantially in the future [2], a trend that likely applies to other developed nations.

The causes for amputation are diverse. In the United States, of the 623’000 amputees, about 81% of the amputations were due to vascular diseases, 17% due to trauma, and 2% due to cancer. Of the amputations due to vascular diseases, 71% were with a diabetes comorbidity [2]. In Germany, causes are estimated to be 87% due to vascular diseases, 4% due to trauma, 2% due to infections, 2% due to cancer, and 5% are not classified [4].

The use of prostheses can restore mobility and participation in daily life to some extent, and it can increase quality of life in general [5, 6]. However, most amputees are still severely limited in mobility compared to able-bodied people. In a review of 35 studies assessing the outcomes of lower limb amputations following trauma, Penn-Barwell reported that only 55% of the above-knee amputees were able to walk 500 m or more [7]. Also the risk of falling is high, with lower-limb amputees having a risk of approximately 50% to fall in one year [8].

A major factor for safe ambulation is the functionality of the prosthetic device; it has been shown that microprocessor-controlled knee joints that modulate joint damping can improve walking performance and reduce the risk of stumbling and
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falling compared to conventional mechanical joints [9]. Recently, such microprocessor-controlled devices with improved sensing and control demonstrated various biomechanical advantages compared to previous generation joints with less sophisticated electronics [10]. However, these devices are still limited compared to their physiological counterparts as they can only dissipate energy.

Devices with a strong actuator capable of producing kinetic energy offer more possibilities in how a prosthesis can act and react. Advancing actuator and battery technology makes such devices feasible today (e.g. the PowerKnee by Össur, which is available commercially [11]), but first studies found little benefits compared to variable-damping devices [12,13], and it was hypothesized that “technology as complex as a powered knee prosthesis may not yet be ideal” [12]. As this new technology matures and the capabilities of actuated transfemoral prostheses are further explored, these shortcomings can hopefully be addressed, and mobility and flexibility of prosthesis users could be improved. Especially activities that intrinsically require positive kinetic energy from the knee joint, such as standing up from a chair, or step-over-step stair climbing, may benefit from actuated knee prostheses, if they can be controlled accordingly. Additionally, actuated devices could enable amputees to perform physical activities that would hardly be possible using variable-damping devices, such as sports involving jumping, where high positive power is demanded from both knee joints.

Even with the most potent actuators and batteries, and most lightweight designs, functional improvements can only be made with appropriate control. In physiological gait, varying muscle activations change not only the forces generated by the lower limbs, but also the limb stiffness. Depending on activity and terrain, one might either walk in a relaxed manner, or one might stiffen up joints by engaging antagonistic muscles to increase stability or to absorb impacts. This stiffening can happen subconsciously through activation-dependent changes in intrinsic muscle properties [14,15] or changes in reflex behavior [16]. With the advent of powered prostheses, modulation of the apparent stiffness of a prosthetic leg becomes possible, and is used today [17,18]. Such modulation may allow the adaptability and complexity of unimpaired gait to be at least partly restored by prosthetic devices. However, the optimal stiffness profiles to be used in powered prostheses are unknown, and as a result such control design is usually associated with long tuning procedures.

As an alternative to manual tuning, a controller could mimic facets physiological gait, for example physiological stiffness modulation. However, quantitative descriptions of how knee joint stiffness varies with muscle activation and co-contraction are still missing. The goal of this thesis is to quantify such variation, and to translate the findings to the design and control of transfemoral prostheses, with the vision to enable more versatile prostheses that ultimately improve the quality of life of amputees.
1.2 State of the Art

1.2.1 Definition of Joint Impedance and Joint Stiffness

A general term describing the mechanical properties of a joint is *impedance*. Joint impedance describes the moments generated at a joint in response to external perturbations of posture [19]. For small displacements, the impedance of a human joint can be described in a linear, time-invariant form, by inertial, viscous and elastic components [20]. The elastic component is often referred to as joint stiffness. The viscous and elastic properties of a joint depend on the passive properties of the anatomical structures surrounding the joint, and on the activation-dependent properties of the muscles spanning the joint [21]. During postural conditions, joint viscoelasticity increases with increasing muscle activation [22, 23], a phenomenon thought to be caused largely by the short-range stiffness of muscles acting about the joint [24, 25]. This stiffness is part of the mechanical properties of the joint, which determine joint behavior in the short time before spinal or cerebral control can modulate the force generated by contractile elements in muscle. In a simplified scheme, this situation is depicted in Figure 1.1. A description of muscle mechanics and underlying mechanisms of short-range stiffness can be found in Appendix A.

The discussed mechanical properties determine the initial joint behavior upon interaction with the environment. Around 100ms after onset of an interaction, reflexes, sometimes denoted as afferent feedback loops, can start modulating muscle forces [16, 26]. This changing muscle activation will influence joint behavior directly through muscle forces, but will also change the mechanical properties of the muscle (see Appendix A). The gains of such feedback loops are thought to be altered by cere-

![Figure 1.1: Intrinsic joint viscoelasticity is variable and changes with varying muscle activation. In a simplified model, the contractile elements are stiff one-directional actuators, and their force generation is controlled by spinal and cerebral control, which entail a time delay. The intrinsic elasticity $k$, viscosity $b$ and lower leg inertia $J_{\text{leg}}$ (with mass $m_{\text{leg}}$) are mechanical properties acting on the joint without any delay. Subscript $e$ denotes extensor muscles, and $f$ flexor muscles. Together these elements produce a net moment $\tau$ about the joint.](image-url)
bral control [27, 28], and reflexes depend on the type of interaction (e.g. amplitude, direction and duration of a perturbation [29]). With slower dynamics, volitional control additionally contributes to behavior of the knee joint. Because reflexive action and intrinsic muscle properties are tightly interlinked, it is difficult to clearly distinguish the contribution of these components to joint behavior [16, 30, 31].

In this thesis, the mechanical properties of joint and muscles are considered, including their dependence on muscle activation. However, the source of the muscle activation (be it reflexes or volitional control) is not considered. The stiffness caused by muscle activation will be denoted as active stiffness, and the stiffness when no muscle activation is present will be denoted as passive stiffness. In some literature, joint stiffness is defined as the apparent covariation of joint moment and joint angle profiles during a task [32, 33]. However, the same profiles can be achieved with different amounts of co-contraction, thereby changing the stiffness of the joint while producing the same moment and angle profiles. Therefore, in accordance with recent literature, this covariation of moment and angle is denoted here as quasi-stiffness, which should not be confused with the intrinsic stiffness of the joint [32, 33]. Although definitions and interpretations of stiffness in the literature vary, the common notion is that muscle stiffness plays an important role in human motor control [32].

1.2.2 Identification of Physiological Knee Stiffness

Experiments to determine joint stiffness typically involve perturbing the joint in a controlled manner and measuring the corresponding motions and moments. These methods have been applied to assess passive knee joint stiffness (i.e. when muscles are not activated) [19, 34, 35]; research on active joint stiffness (i.e. joint stiffness attributed to activation-dependent muscle elasticity) for the lower limb mainly focused on the ankle joint, for example investigating dependence of stiffness on ankle angle [36], ankle moment [22, 37], influence of co-contraction [38], or stiffness variation due to non-isometric contractions [39].

Fewer studies have been published on active knee joint stiffness. Zhang et al. considered conditions without co-contraction, at different angles and contraction levels [23]. McHugh and Hogan quantified stiffness at maximum voluntary contraction (MVC) at different flexion angles [40]. Tai and Robinson measured stiffness during co-contraction, but did not control for the amount of co-contraction [41]. Granata et al. compared gender differences in active knee stiffness, using transient perturbations applied manually by human investigators [42]. All these studies have only considered isometric conditions, and none of them have considered controlled amounts of co-contraction, or dynamic conditions such as gait.

Following the assumption that knowledge of physiological impedance could facilitate control design for prostheses, several research groups have recently started working on quantifying physiological joint impedance during gait using special apparatus that apply perturbations, mainly for the ankle joint; ankle stiffness was
1.2. State of the Art

identified at distinct time instances during stance phase of walking [43], and during swing phase [44]. Tucker et al. has developed a device that is able to perturb the knee joint during gait for the purpose of stiffness identification [45], but no results are available yet. However, all these approaches exhibit the same drawbacks: Identification of joint stiffness requires a dedicated apparatus and is limited to exactly the conditions during the measurement, and it is unclear whether the perturbing device does not alter the gait substantially.

In contrast to perturbation-based approaches, model-based approaches can estimate joint stiffness without the need to apply perturbations. One study recently proposed a model for the force-dependent endpoint stiffness variations of the human arm [24], but conditions with co-contraction were not considered. As co-contraction varies heavily during gait [46, 47], it is essential that a model-based approach takes this into account, e.g. by use of electromyograms (EMGs). Some models for the upper limb have incorporated EMG, but they either fit parameters to best match the experimentally determined stiffness [48], or used a musculoskeletal model that was determined for the specific subject and task [49]. To the author’s knowledge, there is no model-based approach to estimate knee stiffness that takes into account co-contraction and that has the possibility to generalize the estimation to various conditions, including common locomotor activities.

1.2.3 Transfemoral Prostheses Hardware

Transfemoral prostheses can be classified into three main categories: the purely mechanical devices, the microprocessor-controlled variable-damping devices, and the powered devices.

The purely mechanical devices rely on mechanics only to achieve stance and swing phase control, and usually require a considerable adaptation of the gait pattern of amputees. The earliest and simplest articulated knee joints had a single axis and no swing phase control. However, it was recognized already in the 1950s that some amount of control of the shank during swing phase could be provided by adding constant friction about the single-axis knee joint. This approach could only provide smooth gait at a single walking speed [50]. This shortcoming could be improved to some extent by increasing the friction towards full knee extension [51]. More advanced designs followed that included hydraulic damping, for example the widespread Mauch SNS (swing-and-stance) hydraulic knee [52], which allows different resistance at different walking speeds through optimization of the hydraulic element. This knee is still available today in a refined version (Mauch Knee by Össur, Reykjavik, Iceland). Other designs replaced hydraulic cylinders with pneumatic cylinders [53], mainly to reduce the higher costs associated with hydraulic components at that time. In 1976, polycentric knee joints were first introduced in the form of four-bar linkage joints, with the goal to tackle the cosmetic problems that amputees with long residual thighs (e.g. patients with a knee disarticulation) had with
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bulky hydraulic components [54]. Soon, the advantages of polycentric joints were recognized also for amputees with shorter residual limbs, as they can provide higher stability in stance phase when the instantaneous center of rotation lies posterior to the shank and closer to the hip joint, and they can increase foot clearance during swing phase [55, 56]. The combination of polycentric knee joint and fluidic damper presents the latest development for purely mechanical devices, sometimes denoted as hybrid prosthetic knee joints, to exploit the advantages of both concepts (e.g. the 3R78 by Otto Bock, Duderstadt, Germany, or the Total Knee 2100 by Össur, Reykjavik, Iceland). Compared to a physiological knee, these prostheses still present a major limitation, as control of the knee joint is still purely governed by external mechanics.

The second category of transfemoral prostheses are microprocessor-controlled variable-damping devices, which aim to improve the possibilities of controlling the prosthesis. They became clinically feasible in the past two decades, thanks to the advent of affordable, power-efficient microelectronics. The first commercially available device that employed a microprocessor-controlled pneumatic damper was the IP (Intelligent Prosthesis, by Chas. A. Blatchford & Sons Ltd, Sheffield, United Kingdom) [57, 58]. It used a proximity switch to detect stance time and altered the level of damping depending on walking speed. The most widespread variable-damping device today is the C-Leg (Otto Bock, Duderstadt, Germany). A knee angle sensor and strain gauges in the shin are used as sensors to modulate viscosity of a hydraulic damper depending on gait phase and gait speed. The RheoKnee, which was originally developed at MIT [59] and later commercialized by Össur (Reykjavik, Iceland) uses the same principle, but employs a magnetorheological instead of a hydraulic damper. Many studies have investigated the benefits of microprocessor-controlled variable-damping devices compared to purely mechanical devices. The risk of falling and stumbling has been reported to significantly decrease with microprocessor-controlled devices [9, 60, 61]. It has been hypothesized that energy expenditure by amputees could also be decreased, but in this respect the advantage of microprocessor-controlled knees is not evident or only visible in certain conditions [62–65]. Recently, a more advanced version of the C-Leg, the Genium (initial codename X2) was introduced by Otto Bock. While the principles employed in this device are still the same as in the original C-Leg, it uses more sensors and new control algorithms. It has been shown to significantly improve the Prosthesis Evaluation Questionnaire scores compared to the C-Leg [66], which represents patient-perceived prosthetic-related function and quality of life. Furthermore, it has been shown to facilitate more natural gait biomechanics compared to the original C-Leg [10]. Although clear improvements can be observed in this latest generation of variable-damping devices, they are still limited in their capabilities by design. As a result, the gait of unilateral transfemoral amputees is asymmetric [67], and tasks where able-bodied subjects heavily rely on positive knee joint power, such
as step-over-step stair climbing, require considerable adaptation of the motor pattern [10].

To address these problems of variable-damping devices, research around a third category of prosthetic devices has evolved: the powered devices. Individual attempts with powered devices have been reported already in the late 1970s [68] and 80s [69], but only recent advances in energy storage and actuator technology have made powered devices clinically feasible. Hence, this topic has not attracted much attention until the last decade, and only few studies have been reported. One powered transfemoral prosthesis is commercially available (the PowerKnee by Össur, Reykjavik, Iceland); but preliminary results show only small improvements over variable-damping devices in a few conditions [12, 13], which has been attributed to the early development stage of powered prostheses [12]. This is why several research groups have proposed new designs for transfemoral prostheses in the past few years. The vast majority uses electric motors [18, 70-73] as actuators, but also a self-contained, powered hydraulic device has been presented, in which the hydraulic piston can also act as a damper [74] (see Figure 1.2). Although various designs have been presented, none of these designs were optimized to replicate physiological stiffness, which may render them inappropriate for controllers that aim to mimic physiological behavior.

![Figure 1.2: Transfemoral prostheses from other research groups, from left to right: University of California, Berkeley, CA, USA [74], Vanderbilt University, Nashville, TN, USA [71], Clarkson University, Potsdam, NY, USA [72] and Massachusetts Institute of Technology, Cambridge, MA, USA [18]. Pictures taken from the respective publications.](image)

### 1.2.4 Transfemoral Prostheses Control

One of the major challenges in the development of electronically-controlled knee joints is appropriate control of the device for different activities, and throughout the gait cycle of a single locomotor activity. Variable-damping knee prostheses almost exclusively use a gait phase detection, which tries to detect the phase in the gait cycle from a predefined number of states, based on sensor signals from the prosthesis [10,
1. **Introduction**

Depending on the estimated phase, a different level of damping is applied. This presents severely limited control possibilities, because the joint can only brake, and makes it difficult to replicate biological function.

Powered prostheses have more possibilities in how they can be controlled: Rather than only brake, they can dissipate and provide power arbitrarily. Early approaches have used for example echo-based control, where the motion of the sound knee is mirrored on the prosthesis with a time delay and is executed in position-control mode [68], something that would not have been possible with variable-damping devices. However, echo-based control limits the use of the device to symmetric activities, and comes with a time delay. Moreover, position control is thought to be far away from physiological control principles, leads to high interaction forces, and can be dangerous for humans.

Control of powered devices may benefit from mimicking physiological control principles, in particular knee stiffness. Such functionality is not implemented in any devices today. Most powered devices use some form of evolution of the gait phase detection known from variable-damping devices. Powered devices cannot only dissipate energy, but they can also generate kinetic energy. This energy is often provided by a virtual stiffness, essentially modeling a rotary spring about the knee joint with a certain setpoint angle (also called equilibrium angle). Depending on the actual angle of the knee with respect to the setpoint, a different joint moment is applied. Often, this virtual stiffness is combined with a virtual damping, which applies moments depending on angular velocity of the prosthesis. Parameters for virtual stiffness and damping are commonly subsumed to virtual impedance, and controllers that switch between different impedance parameters depending on gait phase are called variable-impedance controllers [18,70,71,73,74]. Tuning of the parameters is usually a tedious manual task, but methods are emerging to automate this process to some extent [75-77]. However, it is unknown in how far the obtained impedance profiles reproduce biological function, because quantitative data on physiological knee joint impedance during gait is still missing.

A control approach that could theoretically replicate biological stiffness modulation is EMG-based control. It could leverage the extraordinary control and adaptation capabilities of the human sensory-motor system, by measuring muscle activity of residual muscles to control a prosthetic device. EMG-based approaches are already used in commercial upper limb prostheses (e.g. the Electric Hand from Otto Bock, Duderstadt, Germany). With increasing research efforts on powered prostheses, promising studies on EMG-based control of transfemoral prostheses have been presented recently [72,78,79]. The EMG-signals have been used in different ways for transfemoral prostheses; in variable-damping prostheses, they were used for proportional control of damping of a variable-damping prosthesis [80] or for activity mode detection [81]. In powered prostheses, they were used to improve accuracy of a activity mode and gait phase detection [78], were used for proportional con-
trol of prosthetic knee moment superimposed with moments from a common gait phase detection [72], or were mapped to setpoint angular velocity of an impedance controller with constant stiffness in non-weightbearing activities [79]. Drawbacks of EMG-based approaches are the noisy and highly variable EMG signals, and the difficulty to access the relevant muscles using electrodes; residual limb length and retained muscles vary substantially for individual amputees, and placing electrodes inside the shaft puts further strain on the residual limb. The latter problem can be partially addressed by targeted muscle reinnervation (TMR), originally developed for upper limb prostheses, a method which surgically transfers nerves from residual muscles to alternative muscle sites [82]. After reinnervation, EMG on these better accessible sites can be measured and used for control of prostheses. This method has recently been applied in a lower-limb amputee, with encouraging results and positive subjective feedback from the patient [83]. However, TMR involves a complicated surgery only performed as part of the amputation, and, hence, is not useful for most amputees. Apart from the disadvantages associated with EMGs explained in this paragraph, none of the approaches use EMG to modulate knee joint stiffness, which could be done by measuring the relative activations of antagonistic muscles. Hence, also EMG-based control could benefit from quantitative knowledge of knee joint stiffness during gait.

1.3 Contributions and Outline

The first goal of this thesis is to develop an approach to quantify physiological knee stiffness, as it plays an important role in human motor control (Chapter 2). The second goal is to deduct requirements for the design and control of transfemoral prostheses that can mimic a biological knee not only in terms of moment generation and velocity, but also in terms of joint stiffness (Chapter 3). The third goal is to design an actuator based on these requirements (Chapter 4). The fourth goal is to evaluate the feasibility of using the actuator in a prosthesis prototype (Chapter 5).

Chapter 2 presents the first step in obtaining estimates of knee stiffness modulation during gait. Specifically, an EMG-guided, model-based approach for estimating knee stiffness is developed. This approach is validated during isometric contractions with substantial co-activation of antagonistic muscles, by comparing it to experimental estimates of knee stiffness. The experimental measurements consider a range of perturbation amplitudes, so as to identify the range over which model-based predictions would be valid. A sensitivity analysis is used to evaluate the influence of model parameters on the accuracy of the estimated knee stiffness. Chapter 2 has been published as a paper in IEEE Transactions on Biomedical Engineering and was used in this thesis with minor modifications (©2012 IEEE. Reused with permission, from [84]).

Chapter 3 uses the developed estimation method and information from the litera-
1. Introduction

ture to deduct precise requirements for knee prostheses. These requirements include not only moment generation and velocity, but also joint stiffness. They are based on the capabilities of a physiological knee. Requirements for moment generation and velocity are obtained from physiological data from the literature. For knee stiffness, experiments in a gait lab are analyzed to obtain stiffness estimates during gait. To the author’s knowledge, these estimates are the first quantitative estimates of active knee stiffness during gait. Due to large variations in moment arms, the sensitivity to uncertainties in muscle moment arms over the whole range of motion is also investigated. To obtain precise requirements for a prosthetic joint, the full stiffness capabilities of a physiological knee in function of joint angle are analyzed. Chapter 3 reuses material that has been published in conference contributions (©2011 and 2012 IEEE. Reused with permission, from [85, 86]).

In Chapter 4, the requirements obtained from physiology – in particular joint stiffness – are used to design an exemplary prototype that aims to approximate those requirements and that can be used to test different control strategies. The possibility of combining a polycentric knee joint with an actuator to optimize the angle-dependent transmission ratio is explored. Finally, a new variant of Series (Visco-)Elastic Actuator (SVA) is proposed, where a nonlinear transmission changes the actuator stiffness in function of joint angle, optimized for the stiffness-angle profile of a physiological knee joint. The device is named ANGELAA (ANGle-dependent ELAstic Actuator). Instead of metal springs, which are commonly used as series-elastic elements in such devices, the concept relies on rubber cords. In addition, the concept also incorporates parallel springs that were positioned such that in combination with the actuator, the asymmetric moment-angle profile of a physiological joint is approximated. An observer accurately estimates force from deflection despite hysteresis in the rubber. The performance of the device is evaluated in a test setup. To the author’s knowledge, this is the first device optimized specifically for the replication of physiological stiffness. Parts of Chapter 4 have been published in a conference contribution (©2012 IEEE. Reused with permission, from [86]), and major parts have been published online in the IEEE Transactions on Mechatronics (©2014 IEEE. Reused with permission, from [87]), and are to be published in print.

To experimentally assess whether the actuator can be used in a transfemoral prosthesis to restore near-natural gait, it is tested with an amputee subject. The methods and findings are reported in Chapter 5. A control scheme is developed that makes use of the estimates of stiffness during gait obtained in Chapter 3. It combines these estimates with an existing control approach that infers motion of the prosthesis from residual body motion. For comparison, conventional state-machine control is implemented as commonly proposed in the literature. Finally, these control schemes are tested in treadmill walking, and compared to the subject’s own prosthesis, one of the latest variable-damping devices. The collected biomechanical data are compared to data obtained from an able-bodied subject.
Physiological Knee Stiffness in Isometric Contractions

2.1 Introduction

During unimpaired gait, humans vary not only the forces generated by their lower limbs, but also the limb stiffness (see Section 1.2.1). Depending on the terrain, one might either walk in a relaxed manner, or one might stiffen up joints by engaging antagonistic muscles to increase stability or to absorb impacts. This stiffening can happen subconsciously through activation-dependent changes in intrinsic muscle properties [14,15] or changes in reflex behavior [16]. Lower-limb amputees, who rely on prostheses do not have this ability, and therefore are often challenged when locomoting across different terrain. With the advent of powered prostheses, modulation of the apparent stiffness of a prosthetic leg is now possible [17,18]. Such modulation may allow the adaptability and complexity of unimpaired gait to be replicated in prosthetic devices. First, however, more quantitative descriptions of the physiological variations in leg stiffness are required.

Few experimental estimates of knee stiffness have been reported. McHugh and Hogan quantified knee stiffness during maximal voluntary contractions at different flexion angles [40]. The most extensive study assessing knee stiffness was conducted by Zhang et al., who quantified the increases in knee stiffness with changes in isometric flexion and extension moments [23]. Granata et al. also found an increase of knee stiffness with increasing knee moments [42]. None of these studies considered a controlled amount co-contraction of agonist and antagonist muscles, but it is known that co-contraction is present during many phases of locomotion [46,47]. The effect of co-contraction on knee joint stiffness has only been addressed by one study [41], in which three subjects were asked to exert an unspecified amount of constant co-contraction. The resulting stiffness was higher than in a relaxed condi-
2. Physiological Knee Stiffness in Isometric Contractions

While efforts have been made to quantify ankle stiffness during locomotion using perturbations [43, 44], direct estimation of knee stiffness during locomotion has not yet been reported, though a device for that purpose has recently been developed [45]. The lack of data for experimental knee stiffness during gait is likely due to challenges associated with making such measurements. These include the need to attach rigidly to the limb so that appropriate perturbations can be applied, while simultaneously not impeding natural behavior. Some have attempted to estimate knee stiffness by examining the slope of the moment angle curve [88], but this represents the quasi-stiffness and not the instantaneous mechanical properties of the joint [32, 33].

An alternative approach to direct experimental estimates is to obtain model-based estimates. It was recently demonstrated that the stiffness of the human arm can be accurately estimated during isometric tasks by considering only the geometry of the limb and the short-range stiffness properties of the active muscles [24]. While these results were encouraging, they only considered conditions in which minimal co-contraction was present. A critical step in the model-based approach is the estimation of muscle forces. During tasks in which there is little co-contraction, muscle forces can be estimated reliably by considering the forces generated by the limb, and using optimization to distribute those forces across the relevant muscles [89, 90]. However, such optimization strategies fail in the presence of co-contraction [91, 92]. In contrast, EMG-guided optimization can be used to estimate muscle forces during tasks that involve co-contraction [93, 94]. This hybrid experimental-modeling approach may also be useful for noninvasively estimating knee stiffness during co-contraction.

One of the goals of this thesis is to develop a model-based estimate of knee stiffness that can be used to estimate stiffness during the dynamic conditions relevant to locomotion. This chapter presents the first step in that process. Specifically, an EMG-guided, model-based approach is developed for estimating knee stiffness during isometric contractions with substantial co-activation of antagonistic muscles. This approach is validated by comparing it to experimental estimates of knee stiffness. The experimental measurements consider a range of perturbation amplitudes, so as to identify the range over which model-based predictions would be valid. A sensitivity analysis is used to evaluate the influence of model parameters on the accuracy of the estimated knee stiffness. The results demonstrate that a model-based approach can accurately predict the activation-dependent changes in knee stiffness that occur during isometric contractions.
2.2 Model

To estimate active knee joint stiffness, an algorithm involving three successive steps was used (Fig. 2.1). First, individual muscle forces were estimated from knee moment or EMG recordings. Second, these forces were used to estimate muscle stiffness. Finally, these muscle stiffnesses were mapped to joint stiffness, considering the individual muscle moment arms.

2.2.1 Musculoskeletal Model

The knee portion of the lower-limb model developed by Arnold et al. was used [95]. This knee model contained the muscles listed in Table 2.1. Each muscle-tendon unit was described by the following parameters: maximal isometric force $f_0^M$, pennation angle $\alpha$, optimal fiber length $l_0^M$ and tendon slack length $l_0^T$. The model also employs a generic Hill-type force-length relationship $f(l)$ [96]. All parameters represented a nominal subject, as described by Arnold et al., and were not modified for the purposes of this study. The outputs of the model were the vectors of moment arms $r(\phi)$ and muscle lengths $l(\phi)$.

2.2.2 Estimation of Muscle Forces

Two methods to estimate muscle forces were used. The first involved static optimization, which is sufficient in situations without co-contraction. The second incorporated EMGs to estimate muscle forces when co-contraction was present.

Muscle Force Estimation from Net Joint Moment

The static optimization used a min-max objective function to distribute muscle forces equally after normalization by their maximum isometric muscle force at their current operating length [97]. Other cost functions have been proposed in the literature.
2. Physiological Knee Stiffness in Isometric Contractions

<table>
<thead>
<tr>
<th>muscles</th>
<th>abbreviation</th>
<th>function</th>
<th>EMG recorded</th>
</tr>
</thead>
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<tr>
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<td></td>
<td>yes</td>
</tr>
<tr>
<td>Vastus medialis</td>
<td>VM</td>
<td>extension</td>
<td>yes</td>
</tr>
<tr>
<td>Vastus lateralis</td>
<td>VL</td>
<td></td>
<td>yes</td>
</tr>
<tr>
<td>Vastus intermedius</td>
<td>VI</td>
<td></td>
<td>no</td>
</tr>
<tr>
<td>Biceps femoris long head</td>
<td>BFL</td>
<td></td>
<td>yes</td>
</tr>
<tr>
<td>Biceps femoris short head</td>
<td>BFS</td>
<td></td>
<td>no</td>
</tr>
<tr>
<td>Semitendinosus</td>
<td>ST</td>
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<td>yes</td>
</tr>
<tr>
<td>Semimembranosus</td>
<td>SM</td>
<td>flexion</td>
<td>no</td>
</tr>
<tr>
<td>Gracilis</td>
<td>GR</td>
<td></td>
<td>no</td>
</tr>
<tr>
<td>Sartorius</td>
<td>SR</td>
<td></td>
<td>no</td>
</tr>
<tr>
<td>Gastrocnemius medialis</td>
<td>GM</td>
<td></td>
<td>yes</td>
</tr>
<tr>
<td>Gastrocnemius lateralis</td>
<td>GL</td>
<td></td>
<td>yes</td>
</tr>
</tbody>
</table>

Table 2.1: Muscles used in the knee stiffness estimation process.

(see [90]), but tend to produce similar muscle force estimates [91]. Preliminary studies also demonstrated that the estimated knee stiffness was not sensitive to the choice of cost function [85]. Hence, the one that is most computationally efficient was chosen.

EMG-Guided Muscle Force Estimation

To estimate muscle force when co-contraction was present, a two-step process was employed. First, EMGs were used to estimate the flexor ($\tau_{\text{flex}}$) and extensor ($\tau_{\text{ext}}$) contributions to the net joint moment $\tau$. Then, static optimization was used to estimate the individual muscle contributions to $\tau_{\text{flex}}$ and $\tau_{\text{ext}}$, as described above.

Estimates of $\tau_{\text{flex}}$ and $\tau_{\text{ext}}$ were obtained using surface-EMGs from the 7 accessible muscles (see Table 2.1), together with the following model:

$$\tau = c^T a = c_{\text{ext}}^T a_{\text{ext}} + c_{\text{flex}}^T a_{\text{flex}} = \tau_{\text{ext}} + \tau_{\text{flex}}.$$  \hspace{1cm} (2.1)

In this model, $a$ is the vector of average rectified EMGs for each isometric contraction and $c$ is the parameter vector relating EMG to moment. The vectors $a_{\text{ext}}, c_{\text{ext}}$ contain only the values for the extensor muscles, and $a_{\text{flex}}, c_{\text{flex}}$ for the flexors. The parameter vector, $c$, was estimated using linear regression, as described in Section 2.3.3.

2.2.3 Model for Musculotendinous Stiffness

Musculotendinous stiffness was modeled as muscle stiffness $k_M$ in series with tendon stiffness $k_T$ (e.g. Morgan et al. [98]):

$$k_{MT} = \frac{k_M k_T}{(k_M + k_T)}.$$
Three different tendon models were evaluated. The first was that proposed by Zajac [99], including the nonlinear toe-region. The Zajac model scales the linear region of the tendon stiffness curve by the maximum isometric force of the muscle. Our second model replaced this scaling with that proposed by Cui et al. [100], based on the geometrical parameters of the tendon. This second model, which also contained a toe-region, was found to be most accurate and was used to generate all results except for the tendon model comparisons in Section 2.4.5. The last tendon model exactly matches that proposed by Cui. It contains the same geometric scaling as the second model, but assumes that the tendon stiffness is linear without any toe-region.

Muscle stiffness was modeled as being proportional to muscle force $f_M$ and inversely proportional to optimal fiber length $l_0^M$ [100, 101]:

$$k_M = \gamma f_M / l_0^M.$$  

2.2.4 Kinematic Mapping to Joint Stiffness

Musculotendinous forces $f_{MT}$ and stiffnesses $k_{MT}$ were mapped to joint stiffness $K_a$ by

$$K_a = \frac{\partial r_T}{\partial \phi} f_{MT} + r_T^T K_{MT} r,$$

(2.2)

where $r$ is the vector of moment arms and $K_{MT}$ is a diagonal matrix containing the musculotendinous stiffnesses, $K_{MT} = \text{diag}(k_{MT})$ (see [102]).

2.3 Evaluation Methods

2.3.1 Experimental Setup

Subjects were seated in a Biodex chair and secured with straps across the torso. Movement of the thigh was restricted by a strap and a cushioned wooden board placed above the edge of the chair (Fig. 2.2). A computer-controlled brushless servomotor (BSM90N-3150AX; Baldor Electric Company, Fort Smith, AR) was used to apply angular position perturbations to the knee joint. Angular feedback was provided by an encoder with an effective resolution of $6.3 \times 10^{-5}$ rad. Forces and moments were measured using a six degree-of-freedom load cell (630N80; JR3, Inc, Woodland, CA), and the motor was controlled in real-time using Matlab xPC. EMGs were acquired using an 8-channel amplifier (AMT-8; Bortec Biomedical, Calgary, Canada) with differential pre-amplifiers (APE-500; Bortec Biomedical). The electrodes were Ag-AgCl, pregelled, bipolar electrodes with a pole spacing of 2.5 cm (FT-007, MVAP Medical Supplies, Newbury Park, CA). All analog signals were passed through suitable anti-aliasing filters before data acquisition. Data were sampled at 5000 Hz for the first three subjects, and at 2500 Hz for the other two.

Each subject’s leg was attached to the motor using a custom-made thermoplastic cast extending from the toes to just below the knee. The knee center of rotation
2. Physiological Knee Stiffness in Isometric Contra ctions

Figure 2.2: Experimental Setup. The subject’s torso and thigh were fixed by straps, the thigh was also secured by a cushioned wooden board placed above the edge of the chair. The servomotor applied stochastic angular position perturbations. A load cell between the crank arm and the cast was used to measure knee moment.

was aligned with the axis of the motor and locked into position at 60° knee flexion. The back rest was set at 85°. This configuration was selected to be consistent with previous studies on knee stiffness and to have a low contribution of passive joint stiffness [23].

2.3.2 Experimental Protocol

Five healthy male subjects (age: 27-31 years, height: 175-193 cm, weight: 71-89 kg) participated in the study. All procedures were approved by the Institutional Review Board at Northwestern University and all subjects gave informed consent. Only the right leg was measured.

Joint impedance describes the forces generated at a joint in response to external perturbations of posture. Knee impedance was estimated using stochastic displacement perturbations with a standard deviation (SD) of 0.5°. The frequency spectrum of the perturbations was tailored for the purpose of stiffness estimation, which is the steady state component of impedance. Gaussian white noise was low-pass filtered with a transfer function having 2 poles at 5Hz and 8 poles at 10Hz. This reduced the moment due to inertia, which dominates the perturbation response at higher frequencies. These perturbations allowed impedance to be estimated between 0-10Hz, while keeping the experimental torques to values within measurement range of the load cell.

EMGs were collected from the 7 muscles acting about the knee that were easily accessible using surface electrodes (See Table 2.1). Before the perturbation measurements, subjects performed at least two MVcs for knee extensors and knee flexors. The obtained torques were recorded and used to scale the target moment levels in
the experiment. The experimental trials lasted for 35 seconds, during which subjects were instructed to produce constant knee moments, and to not react to the imposed perturbation.

There were two experiments, with a short break in between. The first experiment had a constant perturbation amplitude of $0.5^\circ$ SD. The target moment levels were 10, 20, and 30 %MVC in the flexion and extension directions. Measurements also were made in a passive condition, during which subjects were instructed to relax (0 %MVC). Each of these 7 conditions without co-contraction was repeated twice, resulting in 14 trials. Similar trials were conducted with co-contraction. These had net moment levels of -10 %MVC, 0 %MVC and 10 %MVC. Two co-contraction levels were tested for each net moment level. These corresponded to a 5 %MVC (low level) and 10 %MVC (high level) contraction in the antagonist muscle. This resulted in 6 conditions with co-contraction with three repetitions each, yielding 18 trials.

Experimental estimates of stiffness are known to vary with perturbation amplitude. The second experiment was designed to quantify the magnitude of these variations relative to the activation-dependent changes predicted by the model. For this experiment, perturbations of 5 different SDs ($0.5^\circ$, $1^\circ$, $1.5^\circ$, $2^\circ$ and $2.5^\circ$) were used. Only conditions without co-contraction, at -10 %MVC, 0 %MVC and 10 %MVC were considered. Two repetitions were obtained for each combination of perturbation amplitude and contraction level, resulting in 30 trials. For both experiments, the presentation order of the trials was randomized and there was a break of approximately 60 seconds between successive trials.

Subjects were provided with visual feedback to assist with the maintenance of the target net moment and co-contraction level. Net moment (target and actual) was displayed for all conditions. For conditions with co-contraction, the co-contraction level also was displayed. When the target moment was zero or in flexion direction, co-contraction was defined as a weighted sum of the three measured extensor muscles; the weight for each muscle was proportional to its modeled maximum isometric force. This scaling was preferred over the regression model described in Section 2.2.2, because the regression model would have been sensitive to variations in muscle force distribution among synergistic muscles with respect to the calibration trial. When the target moment was in extension, co-contraction was defined as the difference between the weighted sum of the extensor muscle EMGs and net moment measured by the load cell. All feedback signals were low-pass filtered (cut-off frequency at 0.5 Hz) to minimize the influence of the perturbation on the feedback.
2.3.3 Data Analysis

Stiffness Identification from Perturbation Measurements

Knee impedance was estimated by fitting a second-order system to the magnitude of the impedance frequency response function [20]:

\[
\left| \frac{\tau(s)}{\phi(s)} \right| = |Is^2 + Bs + K|.
\]  

(2.3)

The parameter \( I \) denotes inertia, \( B \) viscosity, and the static gain \( K \) corresponds to elasticity or stiffness, which was compared to model estimates. This model considers only the perturbation response of the joint, not the steady-state moments due to the constant level of muscle activation or the contributions of gravity; all steady-state moments were removed prior to system identification. For data fitting, the recorded signals were decimated to 100 Hz. The frequency response function was estimated using Fast Fourier Transforms with windows of 256 samples and an overlap of 128 samples. Position and moment data, and the corresponding coherence function [20] and frequency response function for a typical trial are depicted in Fig. 2.3. The first 5 seconds of each trial were discarded, and only the remaining 30 s were used, to eliminate possible transients at the start of each trial. To quantify how well the parametric fit described the frequency response function (Eq. 2.3), the variance in the moment accounted for (VAF) by the parametric fit was analyzed. Because the model only predicts active stiffness \( K_a \), the passive stiffness \( K_p \), estimated from the 0 %MVC trials, was subtracted from all measurements.

Model-Based Stiffness Estimation

For conditions without co-contraction, the model-based estimates of knee stiffness were described as a continuous function of moment, because the optimization criterion used to estimate muscle forces had a unique solution for any net moment.

For conditions with co-contraction, stiffness only could be estimated at the experimentally measured torques since the estimates of muscle activity depended on the measured EMGs within each experimental trial.

EMGs were band-pass filtered from 20-500Hz using a 4th order, zero-phase digital Butterworth filter. The resulting signal was rectified and averaged to provide a single measure of muscle activity for each trial. The “activity” measured from a passive calibration trial without any joint perturbations was subtracted from that in the active trials in an effort to remove baseline noise.

Separate trials were used for estimating and evaluating the regression parameters describing the static EMG to moment relationship (Eq. 2.1). Of the 32 available trials, 20 were randomly selected for parameter estimation and 12 were used for evaluation. This procedure of random selection and cross-validation was repeated 500 times.
2.3. Evaluation Methods

Figure 2.3: Data from a typical trial: position $\phi$ (a), moment $\tau$ (b), corresponding coherence (c) and FRF and its 2\textsuperscript{nd}-order fit (d). For this trial, the VAF was 86%.

**Perturbation-Evoked Muscle Activity**

The model considers only steady-state muscle activity, but some of this activity may arise from reflex pathways excited by the perturbation. To assess this possibility, the relationship between joint velocity and EMG was quantified for each of the measured muscles, by using the linear, non-parametric impulse response function and the variance it accounted for in the EMG. More complex models [103] did not improve the prediction accuracy.

**Sensitivity Analyses**

The influence of the selected tendon model was evaluated by comparing the knee stiffness estimated when using the three tendon models described in Section 2.2.3.

Muscle moment arms vary substantially across individuals [104]. The influence of moment arm variability was assessed using a Monte Carlo analysis with 500 repetitions. For each repetition, the moment arm for each muscle was selected randomly from a Gaussian distribution with a mean and SD matched to that reported by Buford et al. [104]. Each moment arm was expressed as a percentage of the nominal
moment arm used in the model. On average, this percentage was 22.3 ± 5.1% for extensors and 18.1 ± 4.8% for flexors.

To investigate the effect of incomplete EMG recordings and to assess the importance of each muscle used in the process, muscles were systematically removed from the EMG-guided estimation procedure. The EMG-guided procedure (Section 2.3.3) was performed using all possible subsets of 5 and 6 out of the 7 measured muscles. The magnitude of the change in stiffness was analyzed relative to the case where all 7 muscles were used for the estimation, to demonstrate the change that can be expected when certain EMGs are omitted. Mean and SD over all subjects and all trials were calculated.

2.4 Results

2.4.1 Moment-Based Stiffness Estimates

Model-based estimates of knee stiffness matched the experimental estimates very well during conditions in which there was no intentional co-contraction (Fig. 2.4). Under these experimental conditions, muscle force was estimated from the measured joint moment using optimization (Section 2.2.2). Across all subjects, the model-based predictions of stiffness for each trial were 7.3 ± 14.1% lower than the experimental estimates. The VAF in the moment data by the experimental estimates (Eq. 2.3) was 90 ± 2% during the passive trials, and 87 ± 6% in the active trials without co-contraction.
2.4. Results

The experimental data showed a drop in the coherence at the resonance frequency. The resonance frequency shifted from around 1.5 Hz for the passive trials, up to around 4 Hz for the trials at 30% MVC.

2.4.2 EMG-Guided Stiffness Estimates

When co-contraction was present, EMGs were used to estimate muscle activity in the flexors and extensors, as well as the corresponding torques produced by these muscles. The sum of the EMG-estimated flexion and extension torques provided an estimate of the net moment about the knee. These predictions of net moment based on EMG measurements closely matched the net moment measured by the load cell in all conditions. In conditions without target co-contraction, the mean difference between measured and predicted moment was \(-0.4 \pm 2.5\) Nm over all trials and all subjects (Fig. 2.5a). Conditions with target co-contraction resulted in a mean difference of \(0.3 \pm 1.8\) Nm (Fig. 2.5b).

The EMG-guided predictions of knee stiffness agreed well with the experimentally estimated values in conditions both without and with co-contraction. Experimental values were slightly lower than model predictions. Over all subjects and all trials without target co-contraction (passive trials excluded), the experimental estimates were \(0.01 \pm 24.4\%\) lower than the EMG-guided model-based estimates (Fig. 2.6a). For the trials with co-contraction, the experimental estimates were \(21.0\pm22.3\%\) lower.
2. Physiological Knee Stiffness in Isometric Contra ctions

<table>
<thead>
<tr>
<th>flex. low CC</th>
<th>flex. high CC</th>
<th>ext. low CC</th>
<th>ext. high CC</th>
<th>zero net low CC</th>
<th>zero net high CC</th>
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Figure 2.6: Active stiffness prediction accuracy based on EMG for conditions without (a) and with (b) target co-contraction (CC). The dashed line in both figures represents the unity line. Each data point represents a trial.

than the model-based estimates (Fig. 2.6b). The VAF in the moment data by the experimental estimates (Eq. 2.3) was 80 ± 7% for the trials with high co-contraction, and 84 ± 5% for the trials with low co-contraction.

2.4.3 Dependence of Experimentally Measured Stiffness on Perturbation Amplitude

Experimental estimates of knee stiffness varied with perturbation amplitude. During 10 %MVC contractions in the flexion and extension directions, net stiffness decreased as perturbation amplitude increased. Over the range of tested amplitudes, stiffness decreased by 27.0 ± 17.2 Nm/rad in flexion, and by 53.9 ± 13.2 Nm/rad in extension (Fig. 2.7a). For the same change in perturbation amplitude, passive stiffness decreased by 27.9 ± 5.9 Nm/rad (Fig. 2.7b). These results were not related to a change in the system dynamics. The 2nd-order model fits had a VAF of 89 ± 4% for the smallest perturbations and 93 ± 2% for the largest perturbations. These results suggest that more than half of the amplitude-dependent change in net knee stiffness during extension contractions comes from the passive properties of the joint, and that nearly all of the observed changes during flexion come from these same passive properties (Fig. 2.7c).

2.4.4 Perturbation-Evoked Muscle Activity

There was no substantial relationship between joint velocity and muscle EMG during the passive trials (VAF < 3% for all amplitudes). In contrast, a larger percentage of the EMG variance could be predicted during the active trials. Across all muscles,
2.4. Results

Fig. 2.7: Dependence of identified stiffness on perturbation amplitude. For each trial, the total stiffness was normalized by the average stiffness of the respective condition (passive, flexion moment, extension moment) identified at the lowest amplitude ($K_{0.5^\circ}$), to illustrate the changes relative to the lowest amplitude. The mean values and SD over all subjects and over the following trials are shown: all active trials (a); all passive trials (b); and the difference between the stiffness in the active trials and the passive trials at the same amplitude, normalized by this difference at the lowest amplitude (c), which illustrates the change in active stiffness $K_a$ only.

the VAF ranged from $18\pm14\%$ for the $0.5^\circ$ perturbation amplitude, up to $32\pm20\%$ for the $2.5^\circ$ perturbation amplitude. These results suggest that some of the muscle activity recorded during the active trials could be attributed to reflex mechanisms excited by the continuous, stochastic perturbations.

2.4.5 Model Sensitivity

Knee stiffness predicted by the model varied substantially with the choice of tendon model. Predictions using Zajac’s tendon model were generally higher than experimentally identified stiffness (Fig. 2.8). In extension, predictions were $14.4\pm15.8\%$ higher, in flexion they were $59.8\pm25.0\%$ higher than experimental values. Cui’s tendon model yielded a good agreement between model predictions and experimental data at high moment levels. Overall, predictions were still higher than experimentally identified values; in extension, they were $3.8\pm17.1\%$ higher, in flexion they were $22.4\pm27.4\%$ higher. The adjusted Cui model, incorporating a toe region, produced the most accurate results across all tested conditions. In extension, predictions were $5.9\pm13.3\%$ lower than experimentally identified stiffness, in flexion they were $8.8\pm15.0\%$ lower.

Model predictions of joint stiffness were sensitive to changes in moment arms, as expected due to the quadratic dependence on that parameter (Eq. 2.2). The SD in stiffness obtained from the 500 Monte-Carlo simulations was $22.3\pm1.1\%$ in flexion, and $27.4\pm1.6\%$ in extension, with respect to the mean over the considered moment range from -40 Nm to 60 Nm. Compared to variation of the moment arms (18.1% in flexion and 22.3% in extension), this resulted in a relative sensitivity of
2. Physiological Knee Stiffness in Isometric Contractions

Figure 2.8: Moment-based predictions of active knee stiffness using different tendon models described in the text. The adjusted tendon model (solid black line) is a combination of Zajac’s and Cui’s model, incorporating a toe region for low forces, as used by Zajac, and a stiffness in the linear region that corresponds to Cui’s model. The experimental data are the same as in Fig. 2.4.

approximately 1.2 in both flexion and extension.

The EMG-guided stiffness estimates were sensitive to which EMGs were used. This was assessed by systematically removing individual EMGs and pairs of EMGs. The effect of removing any single muscle on the predicted stiffness was modest, ranging from approximately 2.2-11.9% across all measured muscles. As would be expected, omitting two EMG signals in the estimation process generally had a higher influence on joint stiffness prediction than omitting one. The largest effect was seen when two highly used synergistic muscles were removed, e.g. VM and VL (change of 41.9%), or ST and BFL (change of 29.1%), presumably making it difficult to estimate moment in direction of the synergistic pair. When omitting other combinations of two EMG signals, the average change in predicted stiffness was less than 16%.

2.5 Discussion

Conventional approaches for estimating joint stiffness require perturbing the joint in a controlled manner, usually placing the limb inside a cast. The goal was to develop a less disruptive method that may eventually be applied during functional tasks such as natural gait. Specifically, a model-based method was developed to estimate activation-dependent knee joint stiffness based on measurements of EMG, joint moment and joint angle, and evaluated this approach against experimental measurements during isometric tasks. Performance with substantial co-contraction was
2.5. Discussion

evaluated; co-contraction is common in many functional tasks including locomotion. The EMG-guided stiffness predictions were most accurate when little co-contraction was present, having an error of only 0.01% on average. These errors were comparable to those based only on measures of joint moment (average error of 7% across the same trials). EMG-guided errors increased to approximately 21% when substantial co-contraction was present. These results provide the basis for model-based estimates of knee stiffness that can be used during functional tasks in which experimental measurements are not feasible. They also place bounds on the accuracy that can be expected from such model based approaches.

2.5.1 Experimental Characterization of Knee Stiffness

The experimental estimates of knee stiffness reported here are very similar to values reported by Zhang et al. [23], who considered only conditions without co-contraction. There are no reports on knee joint stiffness during co-contraction that could be directly compared to the results of this study. Tai and Robinson [41] measured stiffness during co-contraction, but did not control for the amount of co-contraction. Nevertheless, their findings of increased stiffness during co-contraction, relative to passive conditions, are consistent with the findings of this chapter. The data further suggests that active stiffness due to the flexors and extensors adds linearly during co-contraction, as would be expected from muscles with minor mechanical interactions [105]. A decrease in VAF was observed for more demanding trials such as trials with co-contraction, indicating that the subjects could not perfectly maintain a constant effort.

If reflex contributions to muscle activation were present in the experiments, they do not appear to have altered the model’s ability to predict knee stiffness. This finding either suggests that reflex contributions were minimal, or that they simply contributed to the steady-state activation of the muscles. In an effort to distinguish between these possibilities, the relationship between the applied perturbations and the measured EMGs was quantified. While there was no substantial relationship during passive trials (VAF below 3%), it was possible to predict 15-30% of the EMG variance during the active trials. This suggests that reflex responses elicited by the small stochastic perturbations used in this study may have contributed to the measured knee stiffness. Given that the model only considers steady-state muscle activations, it appears that any reflex contributions to knee stiffness arose through contributions to the steady state muscle activation. More transient contributions to muscle activation and stiffness, as would arise from non-stochastic perturbations [16], could not be predicted by the model.
2.5.2 Model-based Estimates of Knee Stiffness

To the author's knowledge, this is the first model-based approach to predict active knee joint stiffness. A similar approach was recently used to model the force-dependent endpoint stiffness variations of the human arm [24], but conditions with co-contraction were not considered. Hence, EMG was not used in that previously published estimation process. Some models for the upper limb have incorporated EMG, but they either fit parameters to best match the experimentally determined stiffness [48], or used a musculoskeletal model that was determined for the specific subject and task [49]. In contrast, the current approach was not fit to the experimental data or customized for these specific tasks. While further validation is necessary, especially for dynamic conditions, the approach may be useful for any subject and any situation where kinematic measurements and EMG can be obtained.

The EMG-guided approach tended to overestimate knee stiffness when there was significant co-contraction. This is not surprising given the low levels of co-contraction considered and the fact that the moment-based estimates (Fig. 2.4), which ignore possible antagonist activity, were most accurate at low moment levels. Regardless of the source, these errors point to the limits on the accuracy that can be expected from this model-based approach to stiffness estimation.

2.5.3 Sensitivity of Model Estimates

Muscles are thought to operate in the region of short-range stiffness for fiber length changes of 1-3% [14]. This is often used to explain the decrease in measured stiffness for increasing perturbation amplitude [41,106]. Our results demonstrated that the passive properties of the joint can also contribute to a substantial portion of the observed amplitude dependence. While many studies have focused on the short-range stiffness of actively contracting muscles, this phenomenon also exists in passive muscles (see Proske and Morgan [107] for a review). Furthermore, the short-range stiffness of passive muscles exhibits an amplitude dependence [108,109], as is shown in Figure 2.7. In the active stiffness, almost no amplitude dependent change was observed for flexion. In extension, the change in stiffness from the lowest to the highest tested amplitude corresponded to a change in active stiffness that could be expected from a voluntary contraction change of approximately 5Nm. Hence, at least for the knee, this amplitude dependence is relatively small compared to the large activation dependent changes in knee stiffness.

In contrast to perturbation amplitude, the tendon model had a large effect on the accuracy of the estimated knee stiffness. The tendon model which scaled stiffness properties according to the geometry of the muscle tendon [100] produced more accurate joint stiffness estimates than the model by Zajac, which scaled stiffness properties according to muscle force. By incorporating a nonlinear toe-region at low forces, estimation accuracy at low moment levels could be further increased. These results appear to differ from previous work in the upper limb, demonstrating
2.5. Discussion

that a linear tendon model with Zajac’s scaling parameters for tendon stiffness was sufficient. This apparent difference arises from the fact that the difference between the tendon stiffnesses predicted by the Zajac and Cui methods is smaller for muscles of the upper limb than for those of the lower limb (unpublished results). While the goal of selecting the most appropriate tendon model for the knee was achieved, it is important to reevaluate the applicability of this model for other joints, before employing the same model-based approach for stiffness estimation.

Model-based estimates of knee stiffness also depended on the accuracy of muscle moment arms. These results are similar to previous findings for the arm [24], and can be attributed to the square relationship between muscle moment arms and joint stiffness. The relative sensitivity of stiffness estimates to variations in moment arms was approximately 1.2. Even with this sensitivity, there was a remarkable consistency in the knee stiffness measured across subjects (Fig. 2.4). For applications that require more accuracy, the uncertainty due to moment arm variations may be further reduced by considering population specific, or even subject specific parameters [110].

The number of EMG recordings affected the ability to generate accurate, model-based estimates of knee stiffness. This likely due to the corresponding effect on the estimation of muscle activity during co-contraction. The detrimental effect of reducing the number of EMGs was most dramatic when multiple muscles from a synergistic group were removed from the analysis. For the extensor muscle group, omitting VM and VL in the estimation process had the biggest influence; for the flexor group, omitting ST and BFL yielded the biggest change. For the extensors, these were precisely the muscles that are able produce the biggest force at the investigated joint angle. For the flexors, the ST is not able to produce a very large force, however, its EMG is presumably highly correlated with the neighboring SM [111], the strongest knee flexor at the tested posture. Eliminating any one muscle had a much less dramatic effect, as did the elimination of smaller muscles or muscles not likely used extensively in the tested posture (e.g. the GM and GL). These results suggest that a reduced muscle set could be sufficient for estimating knee stiffness, especially if it provided significant experimental advantages.

2.5.4 Conclusion

A model-based approach was presented for unobtrusively estimating activation-dependent knee stiffness from measurements of EMG, joint angle and moment. It does not require the application of perturbations to the joint, as done conventionally to assess joint stiffness. This indirect method was validated against the more direct perturbation-based approach and found to provide reasonably similar results. Due to the use of EMG, it was possible to quantitatively estimate activation of antagonistic muscles and, therefore, predict joint stiffness during co-contraction, a common feature of many functional tasks. An important feature of this method is that it was not customized to the subjects or fit to any experimental data. Rather, it uses
2. **Physiological Knee Stiffness in Isometric Contractions**

Scalable muscle models that have also been shown to work well for estimating the stiffness of the human arm. As a result, it represents a general approach that should be applicable to joints throughout the body, provided that the musculoskeletal architecture is available.

It should be noted that elastic stiffness is not the only component that defines the mechanical properties of a joint. Viscous and inertial components contribute substantially to joint impedance. This model focuses only on joint stiffness, which is supposed to play a major role in the control of posture and movement [112]. A model for short-range muscle stiffness is used to estimate the stiffness of the knee joint. Given the intended application to a variable-impedance prosthetic knee joint, a similar model for the viscous behavior could be interesting for future work.

The major limitation of the current approach is that it has only been validated for isometric conditions. While intrinsic muscle stiffness has been shown to follow muscle force during dynamic contractions [113], dynamic conditions may also invoke changes in neural commands, such as reflex activations, that are not contained in the current model. While it will be necessary to investigate such possibilities, the current work presents an important step towards quantification of stiffness-modulation during functional tasks, including those with dynamic contractions such as locomotion. Such quantification is important for the development of transfemoral prostheses that aim to replicate those behaviors.
CHAPTER 3

Derivation of Requirements for Biomimetic Knee Prostheses

3.1 Introduction

To allow restoration of function similar to a physiological knee joint, prosthetic devices need to be able to provide sufficient moment, move fast enough, and behave in a similar way in response to imposed perturbations on the joint, such as heel strike. This translates to requirements for the design of a prosthesis in terms of moment and velocity generation, and joint stiffness, which are considered to be the key functional determinants in combination with weight and mass distribution of the leg. As a physiological joint is to be replaced, such requirements for the artificial joint are usually obtained by analyzing the physiological counterpart. Moment and velocity generation are commonly considered in the design of powered prostheses. To our knowledge, physiological joint stiffness has not been considered for the design of such devices, which is the main contribution of this chapter (Sections 3.4 and 3.5).

To allow the replication of the estimated physiological stiffness templates, a prosthetic device needs to be able to modulate its apparent stiffness over a wide range. Existing knee prostheses mostly use stiff, high-bandwidth actuators [70–72]. While stiff actuators are ideal for high-stiffness applications, because of their high bandwidth and high positioning precision [114], a compliant element in series with the motor improves force control performance to allow rendering of low impedances, protects the transmission mechanically from impacts, makes devices safer to interact with humans, and has the potential for energy storage [115], properties that are all desired in prosthetic applications. Actuator units that can vary their physical stiffness combine the advantages of stiff and compliant actuation [116–119], and have therefore already been used in a knee prosthesis [18]. However, mechanisms to arbitrarily vary physical stiffness require a second actuator, and, hence, increase weight
3. Derivation of Requirements for Biomimetic Knee Prostheses

and complexity, which is undesired in prosthetic applications. Furthermore, none of these actuators were designed to replicate intrinsic stiffness profiles of a biological joint.

In Chapter 2, a model-based method was presented to estimate physiological knee stiffness only from conventional kinematic, kinetic, and electrophysiological recordings [84]. This method allows to estimate stiffness profiles in different activities, for example level-ground gait, stair ascent, or stair descent, without applying perturbations. In this chapter, it is shown how this method can be applied to such activities in the example of level-ground walking (Subsection 3.4).

The requirements for a prosthetic knee joint are often deduced from normative gait data from unimpaired subjects (e.g. [120, 121]). Alternatively, assuming a powered knee joint would be used in combination with a passive ankle joint, knee data from below-knee amputees fitted with a passive ankle prosthesis could be used as a reference [122, 123]. During the same activity, the discrepancy in knee moment between unimpaired subjects and below-knee amputees is substantial. For example in stair ascent, peak moment is around 80Nm for unimpaired subjects [121], and 15Nm for below-knee amputees [122]. Hence, it is difficult to deduce the requirements for a powered knee prosthesis from these data, especially when it is unclear whether the knee will be used in combination with a passive or a powered ankle. To obtain more general requirements that are independent of a specific activity or prosthetic foot, the peak performance characteristics of a physiological knee joint were analyzed. This is done for moment and velocity generation based on models from the literature (Section 3.3), as commonly done for powered prostheses. Using our model-based approach to estimate physiological knee stiffness, stiffness trajectories during gait are calculated (Section 3.4), as well as peak physiological capabilities of the knee joint (Section 3.5). This enables comparison of requirements for gait and for more general activities. For a comprehensive overview, kinematics of a physiological knee joint are briefly discussed first (Section 3.2).

3.2 Physiological Knee Joint Kinematics

The physiological knee joint has more than one degrees of freedom. The tibia can rotate about its longitudinal axis with respect to the femur (internal/external rotation). The range of motion for this movement is relatively small near full extension (around 10° [124, 125]), and increases with increasing knee flexion (to approximately 30° when the knee is flexed at a right angle [124, 125]). Furthermore, elasticity of the collateral ligaments allows adduction/abduction in the knee joint of up to ±8° [126] with high loads. However, such high excursions are only reached when these degrees of freedom are subjected to specifically high loads; in physiological level-ground walking, internal rotation is approximately ±5°, and adduction/adduction angles are similarly small, as analyzed by Lafortune et al. using bone pins [127]. While
these small movements might be important to analyze internal loading in the knee joint on different structures, and to design appropriate endoprostheses [128], these movements are very small with respect to the movements along main axis of the knee joint. Therefore, they are neglected for our purpose of designing a transfemoral prosthesis.

However, the degree of freedom which is considered, the flexion/extension movement, does not happen on a single-axis joint in the physiological knee joint. As the knee flexes, the tibial condyles roll and slide back on the femoral condyles, and the instantaneous center of rotation (ICR) changes with knee angle [128,129]. It is hypothesized that this mechanism minimizes the wear on the menisci and the forces acting on the joint surfaces [130]. This polycentric behavior can be approximated by a four-bar linkage, in which the cruciate ligaments form two of the bars [131]. The movement of tibia with respect to femur is depicted in Figure 3.1. In Section 4.3.4, the potential benefits of such kinematics in powered knee prostheses are analyzed and compared to existing, passive, polycentric knee joints.

![Figure 3.1: The polycentric behavior of a physiological knee joint (sagittal view). The trajectory of the instantaneous center of rotation (ICR) is depicted from 0° to 120° flexion, with black dots in intervals of 20°.](image)

The range of motion of a physiological knee joint in flexion/extension is reported
3. Derivation of Requirements for Biomimetic Knee Prostheses

to be approximately 140-160° [132,133]. In common daily-life activities most relevant
for amputees in Western cultures, the whole range of motion is not needed. During
level-walking, maximum knee flexion is around 60° [121,134], it reaches up to 105°
in stair ascent and descent [121,135], and when sitting on a chair it is approximately
90° [136]. However, in non-Western cultures, the full range of motion of the knee
joint is often utilized, due to kneeling or sitting with crossed legs [137,138].

3.3 Physiological Knee Joint Moment, Velocity and
Power

As the goal was to obtain general requirements for a biomimetic transfemoral prosthesis, independent of a specific activity, the peak performance characteristics of a physiological knee joint were analyzed.

The moment capabilities of a physiological knee joint depend on knee angle and
on knee angular velocity [139,140]. This is due to the changing moment arms about
the joint [104], and the force-length and force-velocity relationship of muscle. The
changing moment arms are caused in part by the polycentric nature of the knee
joint described in the section above. A regression model from the literature was
used that predicts peak knee moment depending on knee angle and velocity [139]. It
was applied to an average subject (body mass 70 kg, height 1.75 m). The resulting
peak extension moment depending on angle and velocity is depicted in Figure 3.2.
Peak flexion moments are lower (shown in Figure 3.3 for isometric conditions).

Figure 3.2: Knee extension moment versus angle versus angular velocity profile based
on a model of physiological knee extension [139], assuming a male between 18 and
25 years, height 1.75 m, mass 70 kg.
3.3. Physiological Knee Joint Moment, Velocity and Power

While the regression model predicts peak moment, it does not predict peak velocity. For peak velocity capabilities, the literature indicates that unloaded peak joint velocity is around 700-800 °/s [141,142], but it is unclear at which angles this can be achieved. Therefore, different locomotor activities were also analyzed, as recorded by Riener et al. [121]. Maximum velocities amounted to 320°/s at 35° flexion angle for level-ground walking, 365°/s at 52° for stair ascent, and 320°/s at 55° for stair descent (Figure 3.4). Due to the unclear dependency of knee angle and peak joint velocity, and the observation that peak velocity in common gait activities occurred at similar angles as peak moments (Fig. 3.3), the velocity requirement did not depend explicitly on knee angle, but aimed to achieve a peak physiological velocity of 700 °/s over the entire range.

For knee joint power requirements, the model by Anderson et al. [139] predicts a peak value of 460 W at a flexion angle of 65°, with 69 Nm at 380°/s. Values reported in the literature generally are highly variable. They range from 280 W [143] to 680 W [144] in dynamometer experiments. A value of 500 W was used, as an average of reported values. For comparison, during physiological stair ascent, peak power amounts to approximately 190 W in stair ascent, and -300 W in stair descent for an average subject [121].
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![Graph showing knee velocity vs. knee flexion angle for different activities]

Figure 3.4: Physiological knee joint angular velocity (absolute values) depending on flexion angle in common locomotor activities.

3.4 Physiological Knee Stiffness During Gait

In Chapter 2, a method was presented to estimate knee joint stiffness from kinematic, kinetic and EMG measurements, without the need for perturbations. A validation was also presented for this approach in isometric conditions including co-contraction. In this section, it is shown how this method can be applied to estimate stiffness during gait. During physiological gait, the amount of co-contraction is substantial and accurate estimation of the muscle forces using load sharing (Subsection 2.4.1) is not practical. Therefore, an EMG-guided approach was used to estimate muscle forces. To find bounds on the accuracy that can be expected from variation in moment arms, moment arms reported in the literature are compared and the resulting effect on joint stiffness estimates is analyzed.

3.4.1 Acquisition of Physiological Reference Gait Data

The first step in the process of stiffness estimation is to obtain measurements for joint angles and joint moments. The exemplary data used for these estimates was obtained during level-ground walking from an able-bodied subject (age 29, weight 72 kg, height 1.84 m) at a self-selected walking speed (3.5 km/h on average). An 8-camera Vicon optical tracking system (Vicon Motion Systems Ltd., Oxford, United Kingdom), and a marker setup and processing described by Wolf et al. [145] were used to obtain joint angles. Marker data were sampled at 200 Hz. Ground reaction forces were measured using three Kistler 9260AA6 force plates (Kistler Holding AG, Winterthur, Switzerland) sampled at 3000 Hz. Joint moments were obtained using inverse dynamics as described by Vaughn et al. [146]. Surface EMG from seven easily accessible muscles (rectus femoris (rf), vastus lateralis (vl), vastus medialis (vm), semitendinosus (st), biceps femoris long head (blh), gastrocnemius medialis
(gm), gastrocnemius lateralis (gl)) was recorded. EMGs were collected using a Noraxon TeleMyo 2400R system (Noraxon USA Inc., Scottsdale, AZ, USA), sampled at 3000Hz after analog low-pass filters (8th order Bessel) with a cut-off frequency of 1000Hz. The recorded EMG was rectified, and the envelope was extracted using a zero-lag digital low-pass filter (4th order Butterworth) with a cut-off frequency of 6Hz. The signals were normalized to values obtained during maximum voluntary contractions (MVC), yielding the EMG-based estimate of muscle activity $a_{\text{EMG}}$. Exemplary data is shown in Figure 3.5.

![Figure 3.5: EMG profiles in level-ground walking for extensor muscles (top) and flexor muscles (bottom). Mean and standard deviation over 30 gait cycles are shown.](image)

### 3.4.2 Stiffness Estimation Considering Varying Joint Angles

In contrast to the isometric conditions considered in the previous chapter, moment arms vary heavily during gait. A simple regression from EMG signals to flexor and extensor moment, as used in Subsection 2.2.2 is not able to take this into account. Hence, an approach was used that takes into account the varying moment arms of the individual muscles. Of the 12 muscles present in the model, seven were measured using EMG as described above. The remaining five muscles in the model spanning the knee joint, which are less easy to access, were estimated in analogy to Barrett.
et al. [94]. These muscles were: vastus intermedius (vi), the biceps femoris short head (bsh), the semimembranosus (sm), the gracilis (gr) and the sartorius (sr). The respective activations were:

\[
\begin{align*}
    a_{vi} &= 0.5 \cdot (a_{vm} + a_{vi}) \\
    a_{bsh} &= a_{blh} \\
    a_{sm} &= a_{st} \\
    a_{gr} &= 0.5 \cdot (a_{blh} + a_{st}) \\
    a_{sr} &= 0.5 \cdot (a_{blh} + a_{st})
\end{align*}
\]

A delay of 40 ms accounted for the delay between EMG signal and muscle force [93]. These estimates for each muscle \( i \) were multiplied by \( f_{M,i}^0 \cdot \cos \alpha_i \) to obtain the muscle force; \( f_{M,i}^0 \) is the muscle’s maximum isometric force and \( \alpha_i \) is its pennation angle (values from the literature [147]).

The estimated muscles forces contain inaccuracies, stemming from the noisy and highly variable nature of EMG signals. To correct for these inaccuracies, the resulting flexion and extension moments were corrected by a method similar to Cholewicki and McGill [148]. It essentially adjusts the individual muscle forces as much as needed such that they produce the measured moment. Because observations in an earlier study showed that distribution among synergistic muscles does not have a big influence on resulting stiffness estimates [85], the individual muscle contributions were summarized to extensor moment contribution \( \tau_{ext} \) and flexor moment contribution \( \tau_{flex} \). These were then modified by minimizing

\[
J = (1 - c_{ext}[k])^2 + (1 - c_{flex}[k])^2
\]

with

\[
c_i[k] = \begin{cases} 
    g_i[k], & g_i[k] \geq 1 \\
    1/g_i[k], & g_i[k] < 1
\end{cases} \quad i = \text{ext, flex}
\]

subject to the constraint

\[
\tau_{\text{meas}}[k] = g_{\text{ext}}[k] \tau_{\text{ext}}[k] + g_{\text{flex}}[k] \tau_{\text{flex}}[k],
\]

where \( \tau_{\text{meas}}[k] \) represents the moment measured using inverse dynamics, and \( g_{\text{ext}}[k] \) and \( g_{\text{flex}}[k] \) represent the correction factors at each sample time \( k \). This ensured that the estimated flexion and extension moments reproduced the moment as determined using inverse dynamics. Exemplary net moment measured using inverse dynamics during level-ground walking, and estimated extension and flexion moments are shown in Figure 3.6.

In a next step, the flexion moment was distributed among the flexor muscles, and the extension moment among the extensor muscles (see Subsection 2.2.2). Using these estimated muscle forces, active joint stiffness was calculated (see Section 2.2.4).
3.4. Physiological Knee Stiffness During Gait

This active stiffness only takes into account stiffness due to muscle activation. To account for passive stiffness due to soft tissue and ligaments, average values reported by Zhang et al. [23] were extracted at different knee flexion angles (0°, 30°, 60° and 90°) and values in between were interpolated. Resulting passive stiffness was relatively constant around 35-45 Nm/rad (Figure 3.7). The sum of active and passive stiffness represents the total joint stiffness.

The estimated knee stiffness reached the maximum (around 450 Nm/rad) shortly after heel-strike, and comparably low stiffness could be observed during swing phase (Figure 3.7). Compared to the active stiffness, the passive stiffness estimated using values reported by Zhang et al. [23] was negligible and almost constant over a gait cycle (Figure 3.7).

For comparison, stiffness was also estimated based on net moment (as described in Subsection 2.2.2). It can be seen that estimates using this approach are much lower near heel strike, where the EMG-guided approach estimates substantial co-contraction, and, therefore, higher joint stiffness (Figure 3.7). This comparison confirms that the moment-based estimation is insufficient during gait, because this occurrence of co-contraction is confirmed by the literature [46,47].

3.4.3 Sensitivity to Moment Arms

As already discussed in Section 2.4.5, stiffness estimates are sensitive to muscle moment arms. In the isometric perturbation experiments, only a knee flexion angle of 60° was investigated. During gait, however, the knee angle varies substantially. To
place bounds on the loss in accuracy stemming from different moment arm profiles, the effect of using a different moment-arm sources on knee stiffness estimates was analyzed.

In the original publication of the lower-limb model by Arnold et al., discrepancies between moment arms predicted by the model and moment arms measured by other researchers were already shown [95]. Especially for extensor muscles, the differences were substantial. These results could be reproduced in this study, showing that moment arms predicted by the model can be as much as twice as high near full extension, when compared to values reported by Buford et al. [104] (Fig. 3.8). The data for the moment arm versus angle curves from Buford et al. were obtained by extraction and interpolation from the graphs in their paper [104]. The stiffness estimation process described in the subsection above was repeated using these data (as shown in Figure 3.8). The resulting estimates of knee stiffness show a large discrepancy shortly after heel-strike around 20% gait cycle, when the knee is almost extended, and extension moments are high (Figure 3.9).
3.4. Physiological Knee Stiffness During Gait

Figure 3.8: Moment arm for extensor muscles in the employed musculoskeletal model [95] and data extracted from a publication from Buford et al. [104].

Figure 3.9: Active stiffness estimated using moment arms from the lower limb model by Arnold et al. [95] and compared to stiffness estimates using moment arms extracted from a publication from Buford et al. [104].
3.5 Physiological Peak Knee Stiffness

The previous section showed how stiffness estimates can be obtained during gait. Following the notion that physiological gait data may not be appropriate as requirements for transfemoral prostheses (see Section 3.1), stiffness requirements are now derived based on peak physiological stiffness. To be able to derive stiffness requirements of prosthetic devices, the actuation concept is important. The proposed actuation concept is based on the idea of series elastic actuators (SEAs), which means the transmission is coupled to the load by elastic elements. With accurate force control, a SEA can be used to render different mechanical properties, for example the physiological knee viscoelasticity (Figure 3.10). To translate physiological knee stiffness to design requirements, the finding was used that the apparent stiffness a conventional SEA can display while maintaining passivity is bounded by the physical stiffness of the elastic elements [149]. At the same time, stiffness of the elastic element should be low, because the benefits of a SEA are most apparent when stiffness is minimal. Therefore, the peak stiffness a physiological knee can exhibit is investigated, and should be replicated as closely as possible by the elastic element in the SEA.

To estimate peak stiffness, both antagonistic muscle groups were assumed to be activated at 50\% of their maximum capability as predicted by the model by Anderson et al. [139], the same model used to define peak physiological knee moments (Fig. 3.3). This activation pattern represents a heavy co-contraction. As found in Section 2.4, stiffness increases nonlinearly with activation, with higher rates of increase at lower activations. Hence, this heavy co-contraction yields a higher total stiffness than peak activation of a synergistic muscle group without co-contraction. An even higher activation was assumed to be irrelevant for any common activities, and would likely only be achievable by humans with difficulty. Studies in the literature showing little co-contraction during activities that require high net moments
3.6 Physiological Eigenfrequency of the Lower Leg

Anthropometric data was used to identify properties of a physiological shank and foot. According to Winter [120], the mass of shank and foot is 0.061 times body mass, which amounts to $m_{\text{leg}} \approx 4.57\,\text{kg}$ for a person who weighs 75 kg. Length, as measured from knee joint to ankle joint, is 0.246 times body height, which results in $l_{\text{leg}} \approx 0.431\,\text{m}$ for a subject who is 1.75 m tall. Using these assumptions, the inertia and the natural frequency were calculated when looking at the shank as a pendulum. This is relevant because a prosthesis should swing in a similar frequency as a shank with no muscle activation present. From tabular data from Winter [120] the radius of gyration for the leg relative to the knee joint is

$$\rho_{\text{knee}} = 0.735 l_{\text{leg}} \approx 0.316\,\text{m},$$

which yields the inertia about the knee joint

$$J_{\text{leg}} = m_{\text{leg}} \rho_{\text{knee}}^2 \approx 0.458\,\text{kgm}^2.$$
3. Derivation of Requirements for Biomimetic Knee Prostheses

The natural frequency can be calculated as

\[ f_{0,\text{leg}} = \frac{1}{2\pi} \sqrt{\frac{r_{\text{CoM}}m_{\text{leg}}g}{I_{\text{knee}}}} \approx 0.805 \text{ Hz}, \]

where \( r_{\text{CoM}} \) is the distance of the center of mass (CoM) from the knee joint (0.261 m as determined from tabular data) and \( g \) is the gravitational constant.

3.7 Summary of Physiological Properties

<table>
<thead>
<tr>
<th>Quantity</th>
<th>Physiological capability</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum stiffness during gait</td>
<td>450 Nm/rad</td>
</tr>
<tr>
<td>Peak physiological stiffness</td>
<td>300-1000 Nm/rad depending on angle (Fig. 3.11)</td>
</tr>
<tr>
<td>Moment</td>
<td>100-200 Nm depending on angle (Fig. 3.3)</td>
</tr>
<tr>
<td>Power</td>
<td>500 W</td>
</tr>
<tr>
<td>Velocity</td>
<td>700°/s</td>
</tr>
<tr>
<td>Range of motion</td>
<td>150°</td>
</tr>
<tr>
<td>Shank &amp; foot mass</td>
<td>4.6 kg</td>
</tr>
<tr>
<td>Shank natural frequency</td>
<td>0.8 Hz</td>
</tr>
</tbody>
</table>

Table 3.1: Summarized peak capabilities of a physiological knee joint and shank based on an average subject (body mass 75 kg, height 1.75 m).

The physiological capabilities and properties relevant when designing a transfemoral prosthesis are summarized in Table 3.1. Note that these represent the peak capabilities and it is likely impossible to achieve the same capabilities in a transfemoral prosthesis with current technology. This issue will be addressed in the next chapter.

3.8 Discussion

In this chapter, the capabilities of a physiological knee joint were analyzed and were quantified such that they can be used to derive requirements for a transfemoral prosthesis.

3.8.1 Knee Kinematics

The ICR in the sagittal plane translates 15-20 mm [129, 151], which is a relatively large compared to the moment arms of the muscles spanning the joint, which are in the range of 20-50 mm [95, 104]. Hence, for the physiological actuation of the joint, this translation is very relevant, and is in part responsible for the strong dependency of knee moment capability on flexion angle. However, the literature suggests that for the relative movement between shank and thigh, a single flexion-extension axis that is properly aligned is sufficient for most purposes [152, 153], with errors as small as
3.8. Discussion

3.4 mm in translation and 2.9 ° in flexion [153]. Therefore, the translation of the ICR seems mainly relevant for the internal mechanics of a physiological knee joint, and does not need to be replicated in a transfemoral prosthesis to achieve near-natural gait kinematics.

3.8.2 Knee Joint Moment, Velocity and Power

Peak joint moments were derived using a regression model from the literature [139]. Other studies investigating peak knee moments have found similar dependencies between knee angle and peak flexion and extension moments [154, 155] and support the validity of the model. It should be noted that peak moments in isometric and concentric contractions were analyzed; in eccentric contractions, knee moments can be higher [139]. In a device such as a powered knee prosthesis, friction in the design will likely also increase moment capabilities in eccentric (dissipative) movements.

Peak velocity was based on unloaded knee extension experiments [141], and may differ for knee flexion. In gait, velocities obtained using data from [121] were lower than other data in the literature in level-ground walking (320 °/s compared to 370-400 °/s [120, 156]), and stair climbing (365 °/s compared to 380 °/s [156]). This could be due to differences in subjects' height, which were smaller in [120, 156]. If peak velocity in unloaded knee extension is considered and achieved in a device, all these velocities are achievable.

For peak knee joint power, values from the literature vary substantially from 280 W [143] to 680 W [144]. We choose an average of reported values of 500 W, which still contains a margin compared to maximum power in common locomotor activities (up to 300 W [121, 135]).

3.8.3 Stiffness Estimation During Gait

Stiffness during gait was only estimated only for one subject, but the results should still be representative, because the underlying measurements agreed well with data from the literature: EMG signals were similar [47, 157] as well as joint moments determined using inverse dynamics [120, 121]. Kinematics are more difficult to compare, because absolute values depend on the definition of the zero position. Our absolute values were similar as data reported by Winter [120], but an offset to data from Rienert et al. could be observed, in which minimum flexion angle is around 10 ° [121], while knee angles in our data reach 0 °. This can likely be explained by different definition of the joint axes or zero positions. A method involving calibration tasks was used to determine the functional knee joint axis accurately [145], and the zero position was defined as the position of the knee in relaxed standing, analog to [145].

Our approach to estimate physiological stiffness, which formed the basis for the formulation of requirements, has so far been validated by comparison to isometric perturbation experiments [84]. It has not yet been validated during gait, because it
is very difficult to apply perturbations to the joint without impeding natural gait. Literature suggests that stiffness during movement could be lower than what would be expected from observations in the static case, for example, in the elbow joint [158]. It has also been observed that joint stiffness decreases during movement onset [39, 159]. Future perturbation experiments during gait (for example with the device by Tucker et al. [45]) will show how well our approach estimates stiffness in locomotor activities. It is well known that reflexes play an important role in physiological gait [28, 160, 161], and it might be necessary to consider such mechanisms separately. Nevertheless, our method presents the first attempt to estimate activation-dependent knee joint stiffness during gait.

Another part that was not fully validated in our model was the estimation of muscle force from EMG signals, which are at the base for estimating short-range stiffness. In Subsection 2.4.2, the method was validated in isometric conditions. During gait, different effects may lead to flawed estimation of muscle forces. EMG signals vary with joint angle as muscles move below the surface electrodes [162], and there is a large variation in peak force versus EMG timing [163]. Our approach corrected estimated muscle forces from EMG such that knee moments from inverse dynamics calculations were matched. While this approach can estimate co-contraction, in contrast to a method relying purely on moment measurements, this does not guarantee that co-contraction is accurately estimated.

Using different sources for estimation of moment arms had a substantial influence on estimated stiffness during gait. Moment arm estimates of the employed musculoskeletal model were substantially higher than values reported by Buford et al. [104]. When high knee moments were active, using moment arms by Buford reduced estimates of knee stiffness by approximately 40%. However, the actual loss in accuracy is expected to be smaller, because stiffness estimates using moment arm estimates from the musculoskeletal model were successfully validated in isometric conditions (Chapter 2).

3.8.4 Peak Physiological Stiffness

Our model-based approach showed that peak active knee stiffness varied substantially with knee angle, similar in shape as the peak moment versus angle curve. A study by Hugh and Hogan with imposed knee stretches at maximum voluntary contraction at different flexion angles supports this observation. They reported considerably higher active knee stiffness at 50° and 70° flexion than at 30° or 90° flexion. [40]. They identified a peak value of 450 Nm/rad, but absolute values cannot be directly compared, because they had a different activation (MVC) than our assumption (50% flexor and 50% extensor activation), and they used relatively large stretches of 1.5°, and large stretches are known to yield lower stiffness values [14].
3.8.5 Conclusion

In this chapter, the capabilities of a physiological knee joint were thoroughly analyzed. They were quantified such that they can be used for requirements of a transfemoral prosthesis. Knee moment capabilities were obtained using a regression model from the literature [139]. Requirements for velocity, power and inertial properties were derived from the literature. These are requirements commonly analyzed for powered prosthesis. Requirements for joint stiffness were also derived. To the author’s knowledge, this is the first attempt to derive stiffness requirements for a prosthetic device based on a detailed analysis of physiological knee stiffness. The requirements for stiffness are derived from peak moment capabilities [139] and our method to estimate physiological knee stiffness (Chapter 2).

The method was also applied to estimate stiffness during gait. This presents the first attempt to estimate activation-dependent knee joint stiffness during gait. These estimates are not used to derive hardware requirements, and are not fully validated, but may be useful as an inspiration for control design.
Design and Evaluation of a Transfemoral Prosthesis Prototype

4.1 Introduction

The previous chapter treated the capabilities of physiological knee to obtain precise requirements for joint moment, velocity, and stiffness for a prosthetic device. This chapter addresses how these requirements can be achieved in a prototypical realization. The device should serve as a research platform for different control strategies for transfemoral prostheses.

As discussed in Subsection 1.2.3, there are three categories of transfemoral prostheses. There purely mechanical devices, microprocessor-controlled variable-damping devices, and powered devices. Mechanical devices do not require any sort of control other than the user behavior. Variable-damping devices are limited in their actions, as they can only dissipate energy. Powered devices offer the most possibilities for control, as they can actively generate kinetic energy at the knee joint. This also allows them to replicate behavior of variable-damping devices, if controllers for such devices were to be tested. Hence, a powered concept was chosen for the research platform.

There has been one commercial powered device available for a few years, the PowerKnee (by Össur, Reykjavík, Iceland). First clinical studies with the device were published recently. They found little benefits compared to variable damping devices, which was attributed to the relatively new technology [12, 13]. Therefore, several research groups are working on powered devices [18, 70–74]. Each of those devices offers unique advantages. Martinez-Villalpando and Herr minimized energy consumption by using two parallel SEAs [18]. Lambrecht and Kazerooni presented a semi-active device that can act as a variable-damping device, thereby using little energy, or act in powered mode, providing positive kinetic energy to the knee.
4. Design and Evaluation of a Transfemoral Prosthesis Prototype

joint [74]. Vanderbilt University is working on a prosthesis with powered knee and ankle [70, 71]. A few designs have been presented as test platforms for different control strategies [72, 73, 164]. However, none of them reported the advantages of employing polycentric joints (see Section 4.3), or was designed specifically to replicate physiological stiffness (see Section 4.4).

First, a suitable actuation principle (Section 4.2) is selected and the advantages of employing a polycentric knee joint are explored (Section 4.3). Based on these considerations, a new variant of Series (Visco-)Elastic Actuator (SVA) is proposed, where a nonlinear transmission changes the actuator stiffness in function of joint angle, optimized for the stiffness-angle profile of a physiological knee joint (Section 4.4). Instead of metal springs, which are commonly used as series-elastic elements in such devices [18, 165], the concept relies on rubber cords. In addition, the concept also incorporates parallel springs that were positioned such that in combination with the actuator, the asymmetric moment-angle profile of a physiological joint is approximated. Finally, the performance of the device is evaluated in a test setup (Section 4.5). To the author’s knowledge, this is the first device specifically designed to replicate physiological knee stiffness, which was determined using the presented approach to estimate physiological knee stiffness (Chapters 2, 3).

4.2 Actuation Principle

4.2.1 Hydraulics, Pneumatics or Electric Motors

Our goal was to build a tethered test platform that could be used to test different control strategies for transfemoral prostheses in a laboratory environment. The most important choice was the actuation concept, and it would govern the rest of the design. Three actuation concepts commonly used in robotics and rehabilitation devices were considered: hydraulic, pneumatic, and electric motors.

When the power supply can be stationary and separated from the prosthesis, as in our laboratory test platform, hydraulic actuation would be feasible. This would allow a very high power-density for the actuated joint, which is connected to a stationary pump and motor through hoses. The first powered prostheses used such an actuation concept [166]. Hydraulics are also used in robotics, where the pump is carried by the robot [167, 168], or in exoskeletons [169]. Using considerable design efforts, pump and motor can even fit into a self-contained transfemoral prosthesis prototype [74]. However, hydraulic components are usually associated with high costs and are difficult to control [170, 171], and bulky hoses are handled with more difficulty than electric cables. Because hydraulic systems commonly operate at very high pressures, safety is a concern as small leaks can create dangerous hypertonic needles [172], and flammable oil can be dangerous. Especially in the setting of a prosthetic system, safety is extremely important. For these reasons, a hydraulic actuation concept was discarded.
4.2. Actuation Principle

In terms of safety, pneumatic actuation would be favorable, and the compressor could be stationary with only the piston on the actuated joint. Furthermore, pneumatic systems are inherently compliant, and special designs of pneumatic actuators are known as artificial muscles [173, 174]. However, these designs are still weak compared to physiological muscle. Pneumatic actuation has been used in a first prototype of the transfemoral prosthesis developed at Vanderbilt University [17], and pneumatics have also been used for the actuation of exoskeletons [175, 176]. But due to the lower pressures used in pneumatics compared to hydraulics, pistons generating enough force need to be bigger, yielding bulkier systems. Furthermore, pneumatic systems are very difficult to control [171, 177]. Control could be facilitated to some extent by placing the servo valves directly on the device next to the pneumatic cylinder, but this would lead to even higher weight. Due to these shortcomings, the idea of pneumatic actuation was discarded.

Electric motors are well-studied in robotic control applications, and with sufficient precaution, such actuators can be very safe. Their energy efficiency is also far superior to fluid powered devices [178]. That is why most wearable devices today, including powered prostheses, use electric motors [18, 70, 72, 179, 180]. The increasing awareness of environmental problems as well as exploding numbers of mobile devices will likely boost research on actuator and battery technology even more. Hence, a concept employing an electric motor was chosen.

An electric motor in combination with a common transmission such as planetary gears, Harmonic Drive units, or ball-screw mechanisms, yields a very stiff actuator. While stiff actuators are ideal for high-stiffness applications, because of their high bandwidth and high positioning precision [114], a compliant element in series with the motor improves force control performance to allow rendering of low impedances, protects the transmission mechanically from impacts, makes devices safer to interact with humans, and has the potential for energy storage [115], properties that are all desired in prosthetic applications. Such an actuator is called series elastic actuator (SEA) [181] and will be employed in our prototype due to its favorable properties.

4.2.2 Motor and Revised Requirements

Human joint movements are low-speed and high-torque applications compared to suitably sized electric motors. This is also the case for our design, in which a motor with high power-density and a weight that was acceptable for a prosthesis was selected: A Maxon EC30 4-pole brushless motor (Maxon Motor AG, Sachseln, Switzerland) with 200 W nominal power output, which should satisfy the power requirements for stair ascent. As the motor can be overloaded for short periods, it should also be capable of providing the necessary negative power in stair descent, without the need for additional dissipative elements. Due to its high power-density, compact dimensions and low weight, this motor is used in many powered prostheses [179, 180, 182]. It only weighs 300 g while providing a nominal output torque of
4. Design and Evaluation of a Transfemoral Prosthesis Prototype

0.13Nm, and a nominal speed of 15800rpm. The motor was combined with a drive (Maxon EPOS3, Maxon Motor AG, Sachseln, Switzerland) and power supply that would allow peak torque of four times the nominal values (0.52Nm). To enable such controlled overloading of the motor outside of its specified range, the motor was modified to incorporate two PT100 temperature sensors in its windings.

Even when allowing this overload, a high transmission ratio is necessary to realize the required joint moments. Commonly, Harmonic Drive gears [178] or ball screw mechanisms are employed to this end [70,179]. A ball screw mechanism was selected due to its better back-driveability and higher efficiency compared to Harmonic Drive gears [178,183]. At a comparable transmission ratio, ball screws also offer a slight weight advantage [184].

However, it also becomes apparent that using this motor (and overload mentioned above), the full physiological capabilities (Table 3.1) could not be achieved. Peak power of 500 W could potentially be achieved for short periods of time, assuming the motor torque could be overloaded 2.5 times its nominal value at nominal velocity. But the peaks for joint velocity and joint moment could not be achieved using a constant transmission ratio. To achieve a peak moment of 200 Nm, a transmission ratio of approximately 385:1 would be needed from motor rotation to joint rotation. In contrast, to be able to achieve the peak joint velocity of 700 °/s, a maximum reduction ratio of 135:1 would be allowed. The requirements for our prototype were adapted as a consequence, in particular, the velocity requirement was relaxed such that common activities such as stair ascent and descent, and level-ground walking would still be possible. For a stiff actuator, the velocity requirements can directly be seen in Figure 3.4. However, as a SEA is employed, the compliant elements are first compressed before high loads can be transmitted to the knee joint. This increases the velocity requirements of the actuator. For a first conservative approximation, the higher velocity required can be calculated as the required moment gradient divided by the stiffness of the joint. For peak required moment gradient approximately 1000Nm/s were assumed, which is what occurs during physiological stair ascent (data from Riener et al. [121]). For the stiffness of the joint, 600Nm/rad were assumed, which is an approximation of the lowest stiffness at joint angles where high velocities are required (see Figures 3.4, 3.11). Using these assumptions, required velocity is approximately 100 °/s higher than without the elastic element. Added to the maximum velocity of 365 °/s occurring during stair ascent (Section 3.3), this is rounded to a new relaxed velocity requirement of 500 °/s. The velocity requirement was implemented as a constraint, by limiting the maximum transmission ratio. This also meant that the peak joint moments of a physiological knee could not be achieved, and peak moments produced by the actuator would be around 100 Nm given the constraint placed on the transmission ratio.
4.2.3 Transmission Dimensioning

To select an appropriate ball screw for a slider-crank configuration (Fig. 4.1), several parameters need to be considered. First, the critical rotational speed of the ball screw must not be exceeded. The lead of the ball-screw must be selected such that in combination with the lever arm about the joint \( r_{\text{eff}} \) in Fig. 4.1, the desired transmission ratio is achieved. Finally, it must withstand the high shock loads that could potentially occur.

![Diagram of inverted slider-crank mechanism](image)

Figure 4.1: Inverted slider-crank mechanism of a knee joint with linear actuation. The distance between the line of action of the linear actuator and the instantaneous center of rotation is denoted by \( r_{\text{eff}} \) (the effective lever arm). The attachment point at the shaft is denoted by \( P \), the attachment at the prosthetic shank by \( Q \).

The nominal speed of the motor is 15800 rpm \( (\omega_{\text{M,n}} = 1655 \text{ rad/s}) \), a rotational speed that is too high for most ball-screws of acceptable size and weight. Hence, an additional transmission is needed. A belt drive was assumed with transmission ratio \( i \), which is chosen such that 80% of the critical speed of the ball-screw \( \omega_{\text{BS,crit}} \) was reached at the nominal speed of the motor:

\[
i = \frac{\omega_{\text{M,n}}}{0.8 \omega_{\text{BS,crit}}}, \tag{4.1}
\]

At any time, the knee joint moment was given by

\[
\tau = \tau_M \frac{2 \pi i}{l} r(\phi) \tag{4.2}
\]

and the angular velocity by

\[
\dot{\phi} = \frac{\omega_M l}{2 \pi i} \frac{1}{r(\phi)}, \tag{4.3}
\]

where \( l \) denotes the lead of the ball-screw, \( \tau_M \) the torque of the motor, \( \omega_M \) the rotational speed of the motor, and \( r(\phi) \) is the effective moment arm about the center of rotation at the knee angle \( \phi \). Losses in the belt transmission or ball-screw were neglected. To select a suitable ball-screw and transmission ratio \( i \), it was first ensured that the maximum joint angular velocity could be achieved. The maximum necessary joint angular velocity was assumed to be 500°/s \( (\dot{\phi}_{\text{max}} = 8.7 \text{ rad/s}) \) (see
4. Design and Evaluation of a Transfemoral Prosthesis Prototype

Section 4.2.2). The maximum linear velocity of the ball-screw depends on the lead \( l \) of the screw:

\[
v_{\text{BS,max}} = \frac{\omega_{\text{M,n}l}}{i}.
\]

This yields an upper bound for the allowed effective moment arm acting about the joint center:

\[
r_{\text{max}} = \frac{v_{\text{BS,max}}}{\dot{\phi}_{\text{max}}} \tag{4.5}
\]

Very small moment arms lead to high forces acting on the ball-screw. To avoid damaging the ball-screw independently from control precautions, a lower bound was placed on the effective lever arm based on the maximum permissible force \( F_{\text{BS,perm}} \) acting on the ball-screw:

\[
r_{\text{min}} = \frac{\tau_{\text{peak}}}{F_{\text{BS,perm}}} \tag{4.6}
\]

where the peak moment \( \tau_{\text{peak}} \) was assumed to be 150 Nm/rad, which corresponds approximately to the peak moment of a 100 kg unimpaired person during stair descent [121]. The selection of the ball-screw was a trade-off between the desires to achieve a large range \( r_{\text{max}} - r_{\text{min}} \) and a low weight. Based on this trade-off, a 12 x 5 ball-screw was selected (diameter 12 mm, lead 5 mm, Eichenberger Gewinde AG, Burg, Switzerland). This yielded a transmission \( i = 3.1 \), \( v_{\text{BS,max}} = 0.42 \text{ m/s} \), \( r_{\text{max}} = 48.5 \text{ mm} \), \( r_{\text{min}} = 12.5 \text{ mm} \). For low velocities, the assumption was that the motor could be overloaded up to four times the nominal torque of \( \tau_{\text{M,n}} = 0.13 \text{ Nm} \) for low velocities. This resulted in a maximum force produced by the ball-screw of \( F_{\text{BS,max}} = 1845 \text{ N} \) (neglecting losses in the transmission). When the instantaneous power of the motor exceeded twice the rated power of \( P_{\text{M,n}} = 200 \text{ W} \), the maximum motor torque was further reduced, depending on the required velocity. This resulted in an available torque for the motor of

\[
\tau_{\text{M,max}} = \min (4\tau_{\text{M,n}}, 2P_{\text{M,n}}/\omega_{\text{M}}),
\]

and the joint angular velocity could always be achieved (see Eq. 4.4, 4.5).

4.3 Kinematics: Exploring Polycentric Joints

4.3.1 Introduction to Polycentric Prostheses

In a polycentric joint, the instantaneous center of rotation changes as the joint angle \( \phi \) changes. For passive transfemoral prostheses, polycentric joints are widely used [60], often implemented as a four-bar linkage. The length of the four bars determines the trajectory of the instantaneous center of rotation. This trajectory is also called the centrode. Centrodes of passive polycentric joints usually differ tremendously from a physiological centrode (Fig. 4.2). The shape of the centrode influences how the motion of the prosthesis can be controlled by the user. For example, when the prosthetic knee is fully extended, the instantaneous center of rotation lies behind and
above the actual knee joint, to increase the stability in stance phase [185]. Other advantages over single-axis joints are improved flexion cosmesis [186], and improved foot clearance [187].

Figure 4.2: Four-bar linkage knee joints. A biomimetic model (a) and a conventional polycentric knee joint used in passive prostheses (b). The trajectory of the instantaneous center of rotation (ICR) is depicted from 0° to 120° flexion, with black dots in intervals of 20°.

In active prostheses, knee motion is controlled by the actuation unit, which is driven by some sort of user intention detection (e.g. Martinez-Villalpando and Herr [18]). This makes the need for a specifically shaped centredo obsolete. Rovetta et al. [188] have shaped the centredo to mimic the physiological knee joint, but an analysis of the advantages was missing. Furthermore, an arbitrary centredo could potentially be shaped to optimize the moment versus angle profile of the actuated joint.

In the following, the aim is to approximate the peak moment versus angle versus angular velocity profile of a physiological knee joint. To achieve that goal, the geometry of a joint with linear actuation was optimized and the effects of employing a single-axis joint and different polycentric joints were investigated.

4.3.2 Geometry Optimization

Based on the assumptions for motor and transmission described in Sections 4.2.2 and 4.2.3, the possibility of employing a polycentric knee joint in an actuated transfemoral prosthesis was explored.
Let us first look at a single-axis joint. A ball-screw transmission driven by an electric motor provides a linear force independent of the current nut position. When it drives an inverted slider-crank mechanism with constant force, it produces a non-linear moment profile at the actuated joint, as the effective lever arm $r_{\text{eff}}$ changes with the joint angle (Fig. 4.1). This moment profile is modulated by the attachment points $P$ and $Q$ of the transmission. These points were optimized to yield a moment profile that is similar to a physiological moment profile. With the assumptions above, the squared difference between the target moment profile $\tau$ [139] (Fig. 3.2) and the moment profile $\tau^*$ that resulted from the different attachment points $P$ and $Q$ was minimized:

$$J(P, Q) = \sum_{i} \sum_{j} (\tau(\phi_i, \dot{\phi}_j) - \tau^*(\phi_i, \dot{\phi}_j))^2, \forall i, j \quad \tau^*(\phi_i, \dot{\phi}_j) < \tau(\phi_i, \dot{\phi}_j). \quad (4.8)$$

A grid optimization was used that contained all combinations of coordinates of $P = (P_x, P_y)$ and $Q = (Q_x, Q_y)$ within the bounds $P_x, Q_x \in [-5 \text{ cm}, 5 \text{ cm}]$, $P_y \in [-3 \text{ cm}, 3 \text{ cm}]$ and $Q_y \in [-30 \text{ cm}, P_y - 5 \text{ cm}]$ in steps of 1 cm. This grid optimization was used because more sophisticated optimization methods would have become computationally too challenging, in particular for the case of a four-bar linkage, as described below, where 8 additional parameters need to be found. The coordinates are given in the global coordinate frame ($x$ is anterior-posterior and $y$ is superior-inferior) when the knee joint is completely extended and the leg standing vertically, with the origin of the coordinate frame at the point where the prosthetic shank and shaft intersect. In the case of a single-axis joint, this origin lies exactly on the joint axis.

Using a linear actuator to drive a polycentric joint yields a more complex behavior in how the transmission ratio changes with joint angle. By choosing appropriate linkage parameters, the effective moment arm $r_{\text{eff}}(\phi)$ can be shaped more flexibly compared to a single-axis joint. To achieve this, the same formalism as Hobson and Torfason [189] was used to optimize a four-bar linkage, which has 8 free parameters ($A_x, A_y, B_x, B_y, a_1, a_2, a_3, a_0$, see Fig. 4.3) and only allows linkages that fulfill the Grashof criterion [190]. By employing such a joint instead of the single-axis joint, the cost function (Eq. 4.8) now depended on 12 parameters. All possible combinations of the parameters were calculated on a discrete grid of 1 cm resolution within the bounds: $A_x \in [-2 \text{ cm}, 2 \text{ cm}]$, $B_x \in [A_x + 1 \text{ cm}, 2 \text{ cm}]$, $A_y, B_y \in [-5 \text{ cm}, 2 \text{ cm}]$, $a_1, a_2, a_3 \in [1 \text{ cm}, 4 \text{ cm}]$, and the the same bounds as above were used for $P_x, P_y, Q_x, Q_y$. These bounds keep the size of the prosthetic knee joint within reasonable limits. The angle $a_0$ was adjusted in steps of $20^\circ$ over the whole range of $360^\circ$.

Apart from this optimization, the effect of a physiological centrole was also investigated (Fig. 4.2a), and the use of a conventional four-bar linkage joint employed
in passive prostheses (Fig. 4.2b). Because the range of the centroid was much higher in the conventional four-bar linkage, it was necessary to increase the upper bound for the allowed effective moment arm $r_{\text{max}}$ (Eq. 4.5). This was done by multiplying $r_{\text{max}}$ by factor three, which can be seen as increasing the lead of the ball-screw by factor three. Due to its inversely proportional dependence on the lead, the maximum achievable linear force was consequently divided by factor three. For the two predefined four-bar linkages (biomimetic and conventional passive prosthesis), the 8 parameters of the linkage were given, which reduced the number of optimized parameters to four, the same amount as with the single-axis joint.

The volume spanned between the surface of the physiological moment profile and the surface resulting from the optimized geometry at angles and velocities where the physiological profile exceeded the capabilities of the prosthetic joint was used to quantify the performance of the different geometries. This value differed from the cost function (Eq. 4.8), where the squared difference is used. The obtained value was then normalized by the total volume spanned by the surface of the physiological moment profile.

![Figure 4.3: Four-bar linkage parameters.](image)

**4.3.3 Moment-Angle-Velocity Profiles Using Polycentric Joints**

The optimization yielded very different geometries (attachment points $P$ and $Q$) for the different joints (Figure 4.4). For the optimized four-bar joint, the ball-screw was placed behind the knee joint (Figure 4.4a). The instantaneous center of rotation traveled from the front at $0^\circ$ flexion to the back of the thigh at $120^\circ$ flexion on an almost horizontal path. With a single-axis joint, the optimal geometry using our criterion placed the ball-screw diagonally from the back of the shaft to the front of the shank (Figure 4.4b). Using a biomimetic four-bar, the resulting placement of the
The moment profiles resulting from the optimized geometries matched the physiological capability for many combinations of knee angle and knee velocity, and even exceeded it for some. However, under our assumptions for the motor and transmission, none of profiles came close to the maximum physiological knee moment at low velocities in the middle of the knee range of motion, or the peak isometric moment (194.7 Nm). The geometry with the optimized four-bar linkage yielded a flatter moment profile (Fig. 4.5a) than the other linkages, and the highest peak moment (89.4 Nm when $\phi = 100^\circ, \dot{\phi} = \{0^\circ/s, \ldots, 250^\circ/s\}$). Much of the total volume spanned by the physiological target profile was reached, only 16.5% could not be achieved by the actuator. The geometry with the single-axis joint yielded a moment profile similar in shape as the physiological profile (Fig. 4.5b), with a peak moment of 82.5 Nm when $\phi = 70^\circ, \dot{\phi} = \{0^\circ/s, \ldots, 250^\circ/s\}$. The volume not achieved by the actuator was higher than for the optimized four-bar linkage (24.2% of the physiological volume). The moment profile resulting from the geometry with the biomimetic four-bar linkage slightly exceeded the moment profile using the single-axis joint (Fig. 4.5c). Its peak was 85.3 Nm when $\phi = 60^\circ, \dot{\phi} = \{0^\circ/s, \ldots, 250^\circ/s\}$. Of the total volume spanned by the target profile, 21.1% could not be achieved using this geometry. The moment profile using the conventional four-bar linkage differed the most from the physiological profile (Fig. 4.5d). Its peak occurred in a very different configuration ($\phi = 10^\circ, \dot{\phi} = \{0^\circ/s, \ldots, 350^\circ/s\}$), and was much lower (61.7 Nm). As much as 62.2% of the volume spanned by the target profile was not reached using this joint and geometry.

<table>
<thead>
<tr>
<th>configuration</th>
<th>$P_x, P_y$</th>
<th>$Q_x, Q_y$</th>
<th>$A_x, A_y$</th>
<th>$B_x, B_y$</th>
<th>$a_2$</th>
<th>$a_3$</th>
<th>$a_4$</th>
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<tbody>
<tr>
<td>optimized four-bar</td>
<td>-30,30</td>
<td>-50,-70</td>
<td>-10,-40</td>
<td>10,-20</td>
<td>40</td>
<td>40</td>
<td>40</td>
<td>180°</td>
</tr>
<tr>
<td>single-axis joint</td>
<td>-40,20</td>
<td>50,-120</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>biomimetic four-bar</td>
<td>30,-30</td>
<td>50,-80</td>
<td>-22.8,-22.7</td>
<td>6.9,15.9</td>
<td>32.2</td>
<td>12.8</td>
<td>29.9</td>
<td>100°</td>
</tr>
<tr>
<td>conventional four-bar</td>
<td>-50,30</td>
<td>50,-230</td>
<td>-20.1,-75.0</td>
<td>19.8,55.0</td>
<td>65.9</td>
<td>29.3</td>
<td>56.0</td>
<td>17.6</td>
</tr>
</tbody>
</table>

Table 4.1: Optimized attachment points and linkage parameters (Fig. 4.4). All parameters are given in mm except $a_0$ (in °).

### 4.3.4 Advantages of Polycentric Joints in Actuated Prostheses

This section investigated how the moment profile of an actuated knee prosthesis can be optimized. For a joint with linear actuation, for example an electric motor with a ball-screw transmission, the moment profile depends on joint angle and joint velocity. It was our hypothesis that a prosthetic knee should approximate the moment profile of a physiological knee. Based on requirements derived in Chapter 3, appropriate components were selected and the geometry was optimized. The attachment of the
linear actuator was optimized for different types of joints. The analysis examined the effects on the moment profile of a single-axis joint, a biomimetic four-bar linkage, a conventional prosthetic four-bar linkage joint, and a four-bar linkage with optimized parameters.

The optimization yielded qualitatively different geometries for the different joint types (Fig. 4.4). While the optimal configurations are reported, it could make sense to select a sub-optimal configuration for practical reasons. For example, the ball-screw needs to provide sufficient travel, which could be problematic for the optimal geometry using a single axis joint (Fig. 4.4a); above the attachment point on the shaft, there is no space for the ball-screw to retract, and as both attachment points approach each other for high flexion angles, the ball-screw could potentially stick out of the shank, which is undesirable with respect to cosmesis. To tackle this problem, one could manually select a more practical geometry out of the ones with similar cost, or one could add additional constraints in the optimization.

Our selection of motor, transmission, and constraints guaranteed that the high knee joint velocities during stair ambulation of unimpaired subjects could be reached for all geometries. But none of the geometries was able to reach the high peak moments a physiological knee can achieve in isometric contractions.

A single-axis joint driven by a linear actuator yielded a moment profile similar in shape as a physiological profile, and given our conservative assumptions, would most likely be suitable in a transfemoral prosthesis. The smaller peak moment that was achieved compared to the optimized four-bar linkage was probably due to the coarse grid used in the optimization, which did not allow the single-axis geometry to exhaust the effective lever arm up to the upper bound that was used. Compared to a biomimetic four-bar linkage, the differences were negligible. Using our assumptions and constraints, a conventional four-bar linkage prosthetic joint with linear actuation did not seem feasible. For an optimized four-bar linkage, the resulting moment profile was almost flat over the whole range of motion (Fig. 4.5a). This indicates an almost constant effective lever arm, which was expected from this geometry, because the instantaneous center of rotation moves backwards as the knee flexes, and the linear actuator also moves backwards (Fig. 4.4a). While a flat moment profile is not physiological, amputee gait patterns differ from physiological patterns, and uniform moment capabilities could be beneficial in some cases. How knee torques of transfemoral amputees with an active prosthesis differ from torques in physiological gait patterns most likely depends on the prosthetic foot used, and the control strategy of the prosthesis. Without extensive experiments investigating these effects, it is difficult to predict whether the additional complexity of a four-bar linkage is justified, and an optimized geometry with a single-axis joint seems suitable to assess these effects.
Figure 4.4: Optimized geometry for knee joints actuated by a ball-screw (dotted lines connecting $P$ and $Q$) for an optimized four-bar linkage (a), a single-axis joint (b), a biomimetic four-bar linkage (c), and a conventional four-bar linkage prosthetic knee joint [185] (d). The prosthetic shaft and shank, and rigidly attached elements are sketched in vertical position at 0° flexion angle (thick solid lines). The shank and rigidly attached elements are also sketched at 60° flexion (thick dashed lines). The trajectory of the instantaneous center of rotation (ICR) is depicted from 0° to 120° flexion, with black dots in intervals of 20°.
Figure 4.5: Moment versus angle versus angular velocity profiles for optimized actuator geometries (Fig. 4.4) depicted as solid surface. The target moment profile (Fig. 3.2) is depicted as transparent mesh for reference.
4. Design and Evaluation of a Transfemoral Prosthesis Prototype

4.4 Design and Control of a Prototype with Biomimetic Stiffness

4.4.1 Introduction

As discussed in Chapter 2, joint stiffness is an important property in human motor control, and it is our goal to replicate this stiffness in a prosthetic prototype. To allow the replication of the estimated physiological stiffness templates, a prosthetic device needs to be able to modulate its apparent stiffness over a wide range. While stiff actuators are ideal for high-stiffness applications, because of their high bandwidth and high positioning precision [114], a compliant element in series with the motor improves force control performance to allow rendering of low impedances, protects the transmission mechanically from impacts, makes devices safer to interact with humans, and has the potential for energy storage [115], properties that are all desired in prosthetic applications.

Actuator units that can vary their physical stiffness combine the advantages of stiff and compliant actuation [116, 117, 119, 191], and have therefore already been used in a knee prosthesis [18]. However, mechanisms to arbitrarily vary physical stiffness require a second actuator, and, hence, increase weight and complexity, which is undesired in prosthetic applications. Furthermore, none of these actuators were designed to replicate intrinsic stiffness profiles of a biological joint.

Therefore, a new variant of Series (Visco-)Elastic Actuator (SVA) is proposed, where a nonlinear transmission changes the actuator stiffness as a function of joint angle. The actuator is named ANGELAA (ANGle-dependent ELAstic Actuator). Because the benefits of a Series Elastic Actuator (SEA) are most apparent when stiffness is minimal (guided by the design constraints), its stiffness is minimized while still being able to replicate physiological behavior, following the explicit nonlinear requirements derived in Chapter 3. Instead of metal springs, which are commonly used as series-elastic elements in such devices [18, 165], the concept relies on rubber cords. In addition, the concept also incorporates parallel springs that were positioned such that in combination with the actuator, the asymmetric moment-angle profile of a physiological joint is approximated. An observer accurately estimates force from deflection despite hysteresis in the rubber. The device should serve as a research platform to test different control strategies.

4.4.2 Optimization for Biomimetic Stiffness Profile

After establishing how geometry can be optimized to approximate a physiological moment profile (Section 4.3), this concept is now extended to include physiological joint stiffness. Based on the findings above, the possibility of using a polycentric joint was discarded. Still, the transmission remains a nonlinear function of joint angle, which is exploited to achieve moment-angle and stiffness-angle profiles that
4.4. Design and Control of a Prototype with Biomimetic Stiffness

resemble physiological properties, which are also nonlinear.

Our proposed actuation concept is based on the idea of Series Elastic Actuators (SEAs), which means that the ball screw is connected to the shank via elastic elements. To translate knee stiffness requirements to design choices, a finding by Vallery et al. was used: The apparent stiffness a conventional SEA can display while maintaining passivity is bounded by the physical stiffness of the elastic elements [149]. Therefore, this physical stiffness, reflected on the knee joint, must be at least as high as the maximum required stiffness.

To meet the space constraints given by the envelope of a physiological shank, two pretensioned elastic elements were aligned with the shank, and connected them to the ball-screw actuator over a rocker (Fig. 4.6). This concept exploits the space inside the outline of a physiological shank. It also allows us to choose the ratio between the lengths of the two rocker lever arms \( r_1 \) and \( r_2 \) (Fig. 4.6) to configure the joint stiffness.

This geometry, in combination with the ball screw attachment points, yields nonlinear moment-angle and stiffness-angle profiles, which can be tailored to mimic the profiles of a physiological knee joint.

All parameters describing the geometry (i.e. the attachment points of the elastic elements, the attachment points of the ball-screw transmission, the length, pivot point, and transmission ratio \( r_1/r_2 \) of the rocker element) are found by minimizing the following cost function:

\[
J = \int_{\phi_{\text{min}}}^{\phi_{\text{max}}} \left[ \tau_e(\phi) - \tau^*_e(\phi) \right]^2 + \left[ \tau_f(\phi) - \tau^*_f(\phi) \right]^2 + \lambda \left[ K(\phi) - K^*(\phi) \right]^2 \, d\phi, \tag{4.9}
\]

where \( \tau, K \) represent peak (targeted) physiological joint moment and stiffness (Figures 3.3 and 3.11), \( \tau^*_e, K^* \) represent the approximated values by the actuator, and subscripts \( e, f \) indicate knee extension and flexion, respectively. The weighting factor was \( \lambda = 0.16 \text{ rad}^2 \), and the integral limits \( \phi_{\text{min}} = 0^\circ \), \( \phi_{\text{max}} = 100^\circ \). The parameters were constrained such that the resulting design roughly fit in the outline of a human shank. Furthermore, the maximum lever arm of the linear actuator with respect to the knee joint was constrained, to ensure that the required joint velocities could be reached (500°/s, see Subsection 4.2.2).

A grid optimization with a resolution of 10 mm was used for the ball-screw attachment points, and 5 mm for the spring attachment points and rocker pivot point. The rocker transmission \( r_1/r_2 \) was varied from 0.1 to 0.4 in steps of 0.05.

The optimization was run with a pair of theoretical linear-elastic extension elements with a stiffness of 10 kN/m and a free length of 100 mm.

The force control performance of conventional actuators, including SEAs, can be improved by adding viscosity [192-194]. One effective way of adding viscosity without the need for additional viscous elements (which would increase overall weight) is to use visco-elastic elements instead of linear-elastic elements in the SEA, turning it
into a Series Visco-Elastic Actuator (SVA). Viscoelastic polymers are also generally lighter than linear-elastic (steel) springs.

Therefore, rubber was used as elastic material in the final design. Two pairs of parallel rubber cords (J. G. Karl Schmidt GmbH & Co. KG, Solingen, Germany) take the role of the two theoretical elastic elements on either side of the rocker. Each cord has a diameter of 12 mm, a length of unstressed rubber of 40 mm, and a maximum deflection of 190% of that length (76 mm). Neglecting hysteresis and nonlinearity, the combined stiffness of a pair of cords is approximately 7.3 kN/m in the main operating range. The pair of cords weighs 73 g, including the metal clamps for fixation. In comparison, a steel extension springs with similar stiffness (7.1 kN/m) and maximum deflection (72 mm), has a length of uncompressed spring body of 92 mm, an outer diameter of 25 mm, and it weighs 142 g (Gutekunst + Co. KG Spring Factories, Metzingen, Germany), so it consumes more space and weighs roughly twice as much as the rubber cords. The disadvantage of these viscoelastic properties in terms of hysteresis can be dealt with using an observer [194], as described in Section 4.4.3, such that force can still be measured accurately.

To exploit the fact that the physiological knee is much stronger in extension than in flexion (Figure 3.3), springs acting in parallel to the actuation unit were added to the design (Figure 4.6). Depending on the location of the spring attachment points (P and Q), the moment-angle profile can be very different (Fig. 4.7). For a set of off-the-shelf extension springs, the geometry was optimized to produce a maximal extension moment near 60° flexion, while having only little contribution near full extension and 90° flexion, frequent postures in daily life. The cost function was

$$J = \lambda (\tau_p^*(60°) - \tau_{60})^2 + (\tau_p^*(0°) - \tau_0)^2 + (\tau_p^*(90°) - \tau_{90})^2,$$

(4.10)

with $\tau_p^*(\phi)$ being the moment produced by the springs, $\lambda = 3$, $\tau_0 = 20$ Nm, $\tau_{60} = 50$ Nm, $\tau_{90} = 0$ Nm. The attachment points were constrained to lie within 60 mm posterior and 50 mm anterior to the shank centerline, and 300 mm inferior and 30 mm superior to the knee joint. A trust-region reflective Newton method yielded the optimized attachment points $P : (37, -16)$ and $Q : (-18, -99)$ (in the coordinate frame of Fig. 4.6), in combination with two springs with a stiffness of 9.1 kN/m each, and free length of 103 mm. This spring combination allows the motor to stay below twice its nominal torque during level-ground walking. As parallel elastic elements the rubber cords were not used, as it would have required additional force sensing to accurately estimate their joint moment contribution. This is in contrast to the series-elastic elements, where additional force information is available through motor current (see Section 4.4.3).

The rocker deflection is measured by a 17-bit absolute encoder (Netzer Precision Motion Sensors Ltd., Misgav, Israel). This angle is used to calculate the deflection of the elastic elements and, hence, the output moment at the knee joint. The same type of encoder is used to measure knee angle. Redundant encoders (14 bit, ams
4.4. Design and Control of a Prototype with Biomimetic Stiffness

Figure 4.6: Conceptual design of the prosthesis with serial (left) and parallel compliant elements (right). Both ideas are incorporated in the prototype (Fig. 4.8).

AG, Unterpremstaetten, Austria) are used on the opposite side for safety, such that a sensor fault could be detected.

The finalized prototype roughly fits in the outline of a human shank (Fig. 4.8). The mass of the device without the foot is 3.4 kg, and a conventional prosthetic foot approximately adds further 0.7 kg. The inertial properties of the device were estimated from the CAD model and the measured mass of the device. Inertia about the knee joint is approximately $0.23 \text{kgm}^2$, the center of gravity is 0.22 m distal from the knee axis, and the mass of the shank and foot (without thigh part and parallel springs) is approximately 3.7 kg. This results in a natural frequency of 0.94 Hz (see Section 3.6 for derivation).

4.4.3 Knee Moment Observer

A prerequisite for accurate force control is a reliable measurement of joint moment. In a conventional SEA, this moment is derived from the deflection of the elastic element, normally a metal spring. However, the rubber cords used here are not ideally elastic, they also have a viscous component. The disadvantages of the viscoelastic elements are hysteresis, caused by a retarded elastic response, as well as creep. To obtain accurate estimates of force despite these two effects, both can be compensated by an observer, as described in Parietti et al. [194]. However, since no creep was observed, neither in experiments with this prototype, nor in another robot that uses the same rubber cords [195], the observer was simplified to only take retarded elastic response into account.

A model similar to Parietti et al. [194] was selected, with a spring and a damper in parallel (a Voigt model), and a spring in series to those two elements. The constitutive equation of the model is

$$\ddot{F} + a\dot{F} = c\dot{\delta} + d\ddot{\delta}, \quad (4.11)$$

where $\delta$ is the deflection of the rocker with respect to the shank, and $F$ is the lumped force of the four rubber cords acting on the rocker. The directions of the forces of the
4. Design and Evaluation of a Transfemoral Prosthesis Prototype

Figure 4.7: Knee extension moment produced by a parallel spring assembly ($K = 20 \text{kN/m}$, $l_0 = 100 \text{mm}$) for four exemplary attachment points. Coordinates $(x, y)$ for the attachment points as defined in Fig. 4.6 with the origin at the knee rotation axes are indicated in mm.

Upper and lower cords vary slightly with respect to the rocker, depending on rocker deflection $\delta$. As this dependence is small and has very little effect on the computed knee moment (roughly 0.5% difference) in the operating range, it was neglected.

In analogy to Parietti et al. [194], the observer takes into account additional information about the motor torque $\tau_m$, which is calculated from the motor current, as measured by the EPOS3 drive. The motor equation of motion is

$$J_m \ddot{\varphi}_m = \tau_m - \gamma \dot{\varphi}_m - Fr(\delta, \phi), \quad (4.12)$$

where $r(\delta, \phi)$ denotes the kinematic mapping from rubber cord force $F$ to motor torque $\tau_m$, which depends on the rocker deflection $\delta$ and knee angle $\phi$, $J_m$ is the lumped inertia of motor and transmission including the rocker, $\dot{\varphi}_m$ and $\ddot{\varphi}_m$ are motor velocity and acceleration respectively, and $\gamma$ is the viscous friction coefficient. After integration, the constitutive equation for the rubber cord force (4.11) is

$$\dot{F} - \dot{F}_0 + a (F - F_0) = c (\dot{\delta} - \dot{\delta}_0) + d (\delta - \delta_0), \quad (4.13)$$

with the index 0 indicating initial values at time $t_0$. To simplify notation and without loss of generality, the initial values are assumed to be zero in the following.

When combining the dynamics of the motor part (4.12) and the viscoelastic part (4.13), the system can be described in state-space representation as

$$\dot{x} = Ax + Bu + w \quad (4.14)$$
$$y = Cx + v. \quad (4.15)$$
4.4. Design and Control of a Prototype with Biomimetic Stiffness

The state vector $x$, the inputs $u$, and output $y$ are:

$$\begin{align*}
  x &= \begin{pmatrix} F - c\delta \\ \varphi_m \\ \dot{\varphi}_m \end{pmatrix}, \\
  u &= \begin{pmatrix} \delta \\ \tau_m \end{pmatrix}, \\
  y &= \begin{pmatrix} \varphi_m \\ \dot{\varphi}_m \end{pmatrix}.
\end{align*}$$

The system matrices are

$$\begin{align*}
  A &= \begin{pmatrix} -a & 0 & 0 \\ 0 & 0 & 1 \\ -r(\delta,\phi) & 0 & -\gamma J_m \end{pmatrix}, \\
  B &= \begin{pmatrix} d - ac & 0 \\ 0 & 0 \\ -c r(\delta,\phi) & 1 J_m \end{pmatrix}, \\
  C &= \begin{pmatrix} 0 & 1 & 0 \\ 0 & 0 & 1 \end{pmatrix}.
\end{align*}$$

The process noise vector $w$ accounts for the dry friction when deflecting the rocker, and for backlash present in the bearings. The measurement noise vector $v$ accounts for uncertainties originating from backlash and encoder resolution. A Kalman filter was designed using the covariance matrices of $w$ and $v$, to obtain an estimate $\hat{x}$ of the state vector $x$. The estimate for the rubber cord force $\hat{F}$ is then given by

$$\hat{F} = \begin{pmatrix} 1 & 0 & 0 \end{pmatrix} \dot{x} + c\delta,$$

Figure 4.8: CAD Drawing of the ANGELAA-leg (right) and picture of finished prototype (top left).
4. Design and Evaluation of a Transfemoral Prosthesis Prototype

and the estimate for the joint moment can be calculated as

\[ \hat{\tau} = \hat{F} i(\delta, \phi), \]  

(4.17)

where \( i(\delta, \phi) \) is the geometric mapping between rubber cord force and joint moment. Detailed geometric mapping derivation can be found in Appendix B.

4.4.4 Communications and Control

All sensor signals are sampled at 1 kHz and collected by an STM32 microcontroller. The signals are communicated at 1 kHz via an RS-485 connection to the xPC Target Realtime computer (Speedgoat GmbH, Liebefeld, Switzerland), which runs the control algorithms at 1 kHz. This allows convenient debugging and fast controller development, which was the goal for this tethered prototype. To control the motor, the Maxon EPOS3 drive was used in current control mode and communicating with xPC Target by EtherCAT at 1 kHz.

With the knee moment estimate \( \hat{\tau} \) (4.17), a PI-controller to control knee moment was implemented. The command torque for the motor is

\[ \tau_m = K_P (\tau_{ref} - \hat{\tau}) + K_I \int (\tau_{ref} - \hat{\tau}) \, dt + \tau_{ff}, \]  

(4.18)

where \( \tau_{ref} \) is the reference knee moment; the controller gains \( K_P \) and \( K_I \) were tuned manually and were set to \( K_P = 0.014, K_I = 0.014 \, s^{-1} \). Anti-windup limits the integral term in (4.18) to 1.5 times the motor nominal torque. The feedforward-term is given by \( \tau_{ff} = \tau_{ref}/i_m(\delta, \phi) \), where \( i_m(\delta, \phi) \) denotes the (configuration-dependent) transmission ratio between motor torque \( \tau_m \) and knee moment \( \tau \).

4.5 Calibration and Evaluation of Sensing and Control

4.5.1 Parameter Identification

The parameters in (4.13) were determined using calibration measurements with a JR3 load cell (JR3 Inc., Woodland, CA, USA). The prosthesis was mounted in horizontal position to eliminate gravitational effects, and it was connected to the load cell with a rope (Fig. 4.9). The force along the rope was measured and multiplied by its lever arm about the joint, to obtain an accurate measure of the moment for calibration and evaluation.

Measurements were performed at seven different knee angles for extension moments, and seven angles for flexion moments, ranging from 10\(^\circ\) to 75\(^\circ\) knee angle each. The motor was driven in current control mode to track sinusoidal reference torque profiles, with frequencies ranging from 0.2 Hz to 5 Hz, and amplitudes from 0 to 1.5 times the rated motor torque. The parameters \( a, c \) and \( d \) in (4.13) were identified using least-squares optimization from 10 of the 14 datasets. The 4 remaining
4.5. Calibration and Evaluation of Sensing and Control

sets (2 for flexion and 2 for extension) were used to evaluate the accuracy of this moment estimation.

The parameters in (4.12) were identified using calibration measurements where the rubber cords were removed from the system, so $F$ was equal to zero and the rocker could move freely, detached from the shank. The motor was commanded to follow a sinusoidal reference position trajectory with increasing frequency. Motor acceleration $\ddot{\phi}_m$ was obtained by numerical derivation and smoothing (with cut-off frequency of 10 Hz) of the motor velocity $\dot{\phi}_m$ measured by the drive. The inertia $J_m$ of the drive train and viscous damping $\gamma$ were identified using least-squares regression.

![Figure 4.9: Setup for moment calibration and evaluation.](image)

4.5.2 Performance Evaluation Protocol

To verify the stiffness-angle relationship of the prosthetic knee joint, the identified rubber cords’ nonlinear visco-elastic model was linearized, to obtain an average estimate for the stiffness over the entire range of knee angles. Then, joint stiffness was determined at the distinct knee angles from the experimental data as follows: From the change in rocker deflection angle $\Delta\delta$, the equivalent knee angle deflection $\Delta\phi$ if the motor had been been blocked was calculated. The JR3 load cell measured the change in knee moment $\Delta\tau$. This yielded a measure of the physical stiffness $K = \Delta\tau/\Delta\phi$ at the corresponding knee angle.

To evaluate force control performance, the device was physically fixed at a knee angle of approximately 50° in horizontal position, similar to Fig. 4.9, but without the load cell and fixed with pretensioned ropes in both directions. A sinusoidal reference knee moment was commanded with a frequency that slowly increased from 0.2 Hz to 20 Hz, with multiple oscillations for each frequency. The measured moment was
then compared to the reference moment in terms of phase lag and amplification in steady state for each frequency. This experiment was repeated at different amplitudes ranging from 5Nm to 30Nm.

All experiments in the test setup were performed without the parallel springs and the prosthetic foot.

To assess the feasibility of using the actuator in a transfemoral prosthesis, a pilot experiment was performed with a unilateral transfemoral amputee. The subject was walking on a treadmill at a self-selected speed with two different intention estimation strategies. One controller implemented a simple switching controller, and one was based on the idea of complementary limb motion estimation (CLME). Details of the experiment and the employed intention estimation controllers are presented in Chapter 5. For comprehensive evaluation of the prototype performance, the results of the low-level moment controller are presented here.

4.5.3 Physiological and Replicated Stiffness and Moment

The physical stiffness of the prosthetic joint as a function of knee angle closely approximates the physiological stiffness over the entire angle range (Fig. 4.10). The maximum physical stiffness of approximately 900 Nm/rad is reached at 55° knee angle, and it decreases towards 200 Nm/rad near full extension and towards 300 Nm/rad for full flexion. Compared to the stiffness of the serial springs, the additional stiffness gained by the parallel springs (not included in the figure) is almost negligible (smaller than 30 Nm/rad for the whole range of motion).

![Figure 4.10: Identified physical stiffness profile and measured stiffness of the prosthetic joint compared to physiological stiffness requirements.](image)

The peak moment characteristics of the chosen hardware configuration (given that the motor can be overloaded to up to four times its nominal torque) are shown in Fig. 4.11. The peak moments that actuator and parallel springs can produce are substantially lower than what a physiological knee is capable of (Fig. 4.11).
The theoretical peak moment that can be reached based on these assumptions is 94 Nm at a knee flexion angle of 53\degree, where the actuator delivers 79 Nm and the parallel spring 15 Nm (an increase of 19\%). The contribution of the parallel springs approaches 0 Nm near full knee extension and knee flexion.

Figure 4.11: Peak moment profile of the prosthetic joint compared to physiological moment requirements, assuming a motor peak torque of 4 times its nominal torque. The parallel springs lead to a net joint moment that is biased in extension direction (solid red line).

4.5.4 Moment Measurement

The nonlinear viscoelastic model (4.13) approximated the actual knee moment with a coefficient of determination ($R^2$) of 0.993. Fig. 4.12 shows two of the four validation measurements, and an impact-like situation where the rope was not under tension at the beginning of the measurement. The identified parameter values of the model were $a = 5.47\, \text{s}^{-1}$, $c = 1365\, \text{N}/\text{rad}$, and $d = 6231\, \text{N}/(\text{s} \cdot \text{rad})$, and the parameters in (4.12) were $J_{\text{m}} = 1.54 \cdot 10^{-5}\, \text{kg} \cdot \text{m}^2$ and $\gamma = 4.20 \cdot 10^{-4}\, \text{kg} \cdot \text{m}^2/\text{s}$. Maximum dry friction was 0.025 Nm. The measurements revealed an average absolute error of 0.73 Nm over the four test data sets. Due to the nonlinearity of the kinematics and the slight nonlinearity of the rubber cords’ force characteristics, the moment resolution depends on the knee angle and on the exerted moment; it ranged from 0.003 Nm at zero knee moment and 60\degree knee angle, to 0.013 Nm at 50 Nm knee moment and 10\degree knee angle.

4.5.5 Bandwidth

The moment bandwidth of the prosthetic joint decreased with increasing amplitude, reaching 9 Hz at 5 Nm, and 5 Hz at 30 Nm (Fig. 4.13).
Figure 4.12: Comparison of moment estimates to moment measurement from a load cell for two typical 2 Hz oscillations (flexion in the left plot, extension in the middle plot), and an impact-like situation where the rope was not under tension in the beginning (right plot). The dash-dotted line represents a simple elastic model that does not consider viscoelastic effects, and the dashed line represents the estimates from the observer (Eq. (4.17)).

4.5.6 Moment Tracking During Gait

The amputee was able to walk with both controllers on a treadmill after very little familiarization time (less than 5 minutes).

The statemachine controller tracked the reference moment accurately throughout most of the gait cycle, with less accurate tracking during swing phase (Fig. 4.14). It should be highlighted that the impact situation at heel strike did not pose any problem for the controlled actuator.

The CLME controller had difficulties to track the reference moment especially in pre-swing (Fig. 4.15). It is apparent that during this time, the motor torque reached saturation (at four times its nominal torque). This saturation is a safety feature implemented in the Maxon EPOS3 drive that limits motor current. Again, the impact at heel strike did not cause any problems.
Figure 4.13: Experimental tracking frequency response of the prosthesis for different moment amplitudes. The data points indicate the steady-state response at the corresponding frequency.
Figure 4.14: Reference knee extension moment and actual knee extension moment as measured by the observer during a typical gait cycle in an amputee experiment with a simple switching controller (state machine). Gait cycle is plotted from toe-off to toe-off, to highlight performance during heel-strike (dashed vertical line).
Figure 4.15: Reference knee extension moment and actual knee extension moment as measured by the observer during a typical gait cycle using CLME control. Here, gait cycle is plotted from heel-strike to heel-strike, to emphasize the performance near toe-off (dashed vertical line). Motor torque (bottom plot) reaches saturation shortly before toe-off.
4.6 Discussion

Our goal was to design a tethered transfemoral prosthesis that can mimic the capabilities of a physiological knee joint in terms of stiffness, moment generation and velocity, so that it can serve as a test platform for control design and validation.

Compared to variable-stiffness actuators, our proposed concept does not include a second actuator to adjust the physical stiffness, but the change of physical stiffness is directly coupled to joint angle. The requirements for the relationship between joint angle and stiffness were based on a precise analysis of physiological stiffness-angle relationships. To our knowledge, ANGELAA is the first device specifically designed to replicate physiological knee stiffness, which was possible using our approach to estimate physiological knee stiffness (Chapter 2).

4.6.1 Fulfilling the Requirements

The optimized geometry for ANGELAA enabled the device to cover the required physical stiffness almost completely (Fig. 4.10). In addition, the placement of the parallel springs was able to augment the moment generated by the motor by almost 20\% at knee angles where high extension moments are needed (Fig. 4.11).

However, the peak joint moments of the human knee joint could not be achieved with the given weight constraints placed on the motor. Nevertheless, the device should be usable for common locomotor abilities such as stair climbing: Data from the literature from able-bodies subjects suggests that peak knee extension moments during stair climbing do not exceed 1.1 Nm/kg [121]. Assuming a transfemoral prosthesis that can ideally replicate physiological gait (including an actuated ankle), our prototype would allow stair climbing for subjects of 85 kg, although our requirements were based on only 75 kg body mass. In the case of a passive foot, required knee moments are likely much lower; it has been shown that transtibial amputees with a passive foot prosthesis – a situation comparable to our configuration of a powered knee with a passive ankle prosthesis – need much lower knee moments during stair ambulation [122] than able-bodied subjects. Therefore, the lower knee moments achieved by our actuator compared to peak capabilities should not limit its usefulness as a research platform in common locomotor activities.

The position of the parallel springs was optimized to increase net extension moment where physiological extension moments are highest, which in turn reduces the achievable peak flexion moments. This reduction should be acceptable, as flexion moments during physiological locomotion are comparably small (Fig. 4.11). In the amputee experiment using state-machine control, this hypothesis was confirmed (Subsection 4.5.6). However, when using CLME control, higher motor torques would have been needed to accurately track the reference knee moment (Fig. 4.15). This could either mean that the simplifications made in the design process were too severe and effects such as friction and transmission dynamics are much more important than...
originally assumed. Or it could mean that the reference knee moment that CLME control provided was in fact not meaningful at the time in the gait cycle where this occurs. Given that actual moment and kinematics are very similar for both control schemes (see Chapter 5), and that reference knee flexion moment is much higher than during physiological gait, the latter hypothesis seems more probable. Hence, the reaching of saturation is likely due to issues with reference moment generation, and is not a problem of the actuator.

The mass of the device is similar to the mass of an adult human leg. It is 4.1 kg (including a passive foot), which corresponds to a foot and shank mass of a 67 kg human [120]. About 20% of the American males aged 20-29 are lighter, and about 10% of the 30-79 year old [196]. The weight is comparable to other powered devices [18, 71, 72], but substantially heavier than variable-damping devices such as the C-Leg by Otto Bock, which weighs around 1.2 kg (without foot). It was not our focus to minimize the weight of this prototype, and there is potential to optimize structure and materials to reduce the weight substantially; the employed mechatronic principles do not prohibit a more lightweight design. However, an active prosthesis will always be heavier than a passive one, and future experiments with patients will show whether the benefits of a powered knee joint outweigh the disadvantage of additional weight. It should be noted that it is unclear which weight is acceptable for a transfemoral prosthesis. While it is often stated that it should be as lightweight as possible due to the delicate interface between residual limb and shaft, which is also our experience with clinical partners, there is evidence in the literature that patients do not necessarily prefer the most lightweight devices, at least in passive knee joints [197, 198].

In terms of length, the current prototype with a high-profile foot can be made as long as necessary by using different aluminum tubes, and as short as 0.44 m (measured from rotation axis to the floor); only 1 percent of the American male population has shorter shanks [199]. For knee-exarticulated amputees, the prosthesis has to be shorter than the physiological shank, in which case, depending on their shank length, a low-profile prosthetic foot would have to be employed.

Our device achieved a bandwidth of approximately 5 Hz for a sinusoidal oscillation with an amplitude of 30 Nm. When defining the physiological moment bandwidth as the frequency range over which 70% of the signal is captured (analog to Au and Herr [179]), the physiological knee moment bandwidth was found to be approximately 4 Hz, where knee torque varied between 0 and 35 Nm (based on data from Riener et al. [121]). Therefore, the device should provide sufficient bandwidth.

4.6.2 Hardware Limitations

The mass of the parallel springs is low (84 g per spring) compared to the mass of the device (3.4 kg). However, if the goal is not to replicate the asymmetric moment capabilities, and a motor with similar energy density can be found, a heavier motor
could be used to achieve higher knee moments. But heavier motors with a power density as high as the current motor are rare; a motor with 50% higher torque output and 50% higher weight would likely be preferable to the use of the parallel springs, if such a motor existed. The parallel springs produce only negligible joint moments near $0^\circ$ and $90^\circ$ knee flexion angle, angles required for standing and sitting respectively. This is highly desirable, because it creates natural equilibria and ensures that the actuator does not need to counteract the parallel springs in these frequent postures.

It should be noted that during level-ground walking of able-bodied subjects, the knee joint only generates little positive kinetic energy [120, 121], and entirely passive joints may be suitable for prosthetic or orthotic devices aiming to restore physiological gait, as investigated by several research groups [200-203]. Nevertheless, it has been shown that powered knees can decrease metabolic energy consumption during level-ground walking [204]. Other tasks such as climbing stairs in a physiological way rely on positive energy from the knee joint, which requires a powered device.

Our prototype does not include an actuated foot, which would likely be necessary to restore a completely natural gait pattern. A powered knee with a passive foot can theoretically bring a transfemoral amputee to the level of a transtibial amputee with a passive foot; transtibial amputees have been shown to have better walking abilities than transfemoral amputees [8]. Another approach would be to have a passive knee and an actuated foot, as pursued in the CYBERLEGs project [205], which might help during level-ground walking, but will likely still limit activities such as stair climbing or standing up from a chair.

4.6.3 Visco-Elastic Elements

In contrast to conventional SEAs, viscoelastic rubber cords were used instead of regular springs. The rubber cords have two main advantages over regular steel springs. First, they provide intrinsic damping, which allows the use of higher controller gains and, hence, more accurate and robust force control [192-194]. Second, they are lighter than steel springs of comparable stiffness, allowing for lighter designs that put less strain on the stump. The commonly stated disadvantages connected with viscoelasticity, in particular hysteresis, did not impede our torque sensing: Experimental evaluation showed that our observer structure allowed precise and accurate estimates of joint moment (roughly 1% of the peak moment of the device).

In our design, the serial compliant elements need to be pretensioned, which comes at the expense of extra spring stroke. The actuator could potentially be made more compact by replacing the rocker and the two compliant elements by a rotary spring. Space requirements therefore compare unfavorably to solutions with rotary springs, which can be more compact (e.g. [184, 206]). However, given that our assembly still fits well into the outline of a physiological shank, this should not be a severe disadvantage.
4.6. Discussion

4.6.4 Polycentric Joints

The possibility of using a four-bar linkage mechanism was also investigated, similar to those used in passive prostheses. While they have already been used in active knee prostheses, their designs aimed to mimic a physiological centrodome [188, 207], and the consequences it has for the actuated prosthesis were not analyzed. Section 4.3 presented a detailed analysis of the effects on the moment profile of such a biomimetic joint with linear actuation, and compared it other designs including different polycentric joints. The conclusion was that the increased complexity did not justify the use of a polycentric joint, at least when the goal is to mimic physiological moment-angle profiles; the shape of physiological moment-angle and stiffness-angle profiles can be approximated well using a single-axis joint.

4.6.5 Conclusion

In this proof-of-concept, a new variant of Series-(Visco-)Elastic Actuator was presented that changes its physical stiffness in function of kinematic configuration, optimized to mimic physiology. The concept was used in a prototype of a transfemoral prosthesis (the ANGELAA-leg), and the required stiffness (and moment) profiles were derived from precise analysis of physiological capabilities. The concept contains two further measures to reduce weight: rubber cords as series-elastic elements, and nonlinear parallel elasticity to reduce motor torque requirements. The mechatronic principles described in this paper can also be applied to other impedance-controlled devices where stiffness requirements can be specified in advance and formulated in function of kinematic configuration, and where weight, cost, or power consumption are critical. Examples are other wearable devices like ankle prostheses or leg exoskeletons, but also mobile manipulators.
Case Study

5.1 Introduction

The previous chapters treated detailed requirements for a transfemoral prosthesis based on physiology, and actuator design based on those requirements. To assess whether the actuator could enable near-natural gait with a transfemoral prosthesis, and to compare it to a state-of-the-art variable-damping prosthesis, two control strategies were tested in a case study.

Powered transfemoral prostheses are commonly controlled using state-machine control that switches control parameters based the detected gait phase. The control parameters are usually represented as a virtual impedance, and impedance parameters and switching conditions are obtained using manual tuning and involve 3-4 impedance parameters in 3-5 states [71, 204]. Approaches to automate the tuning process to some extent have been proposed [75–77], but none of them is based on estimates of physiological stiffness.

Recently, a control strategy has been proposed that generates prosthesis command signals from residual body motion without the need for a finite number of states [164]. It is called complementary limb motion estimation (CLME) and has been developed originally to generate reference trajectories for gait rehabilitation robots [208]. It exploits the high interjoint couplings in physiological gait, which are extracted from able-bodied subjects to obtain coupling parameters that are then used for control of the prosthesis. CLME operates in position-control mode, which leads to stiff movements and high impact forces that are uncomfortable and detrimental to the hardware. Therefore, the goal was to improve the control strategy by combining it with the stiffness estimated using the model-based approach presented in Section 3.4. Stiffness reference trajectories are obtained in analogy to position – they are also inferred from residual body motion.
5. Case Study

In this feasibility study, the refined CLME method including physiological stiffness estimates is applied to control the ANGELAA-leg (ANGle-dependent ELAs-tic Actuator). It is compared to a conventional method to modulate stiffness using state-machine control on the same hardware. Both control schemes are tested in treadmill walking and compared to the subject's daily-life prosthesis, a recent variable-damping device (the Genium from Otto Bock, Duderstadt, Germany).

The amputee subject was a unilateral transfemoral amputee (age 44, height 186 cm, weight 75 kg). To analyze the gait of the amputee, a gait analysis involving motion capture was performed to obtain precise kinematic measurements. To evaluate if the amputee gait was similar to physiological gait, the experimental data were compared to data from an able-bodied subject obtained using the same experimental methods. The experiment was approved by the Ethics Committee of the Canton of Zurich.

5.2 Experimental Platforms

5.2.1 Custom Hardware

The prosthesis prototype ANGELAA including electronics is described in Chapter 4. The prosthesis was protected by a custom carbon-fiber cover (Figure 5.3). In one of the experimental conditions, custom goniometers were used to measure hip and knee angles of the sound limb, and hip angle of the amputated limb (Figure 5.1). Details on the goniometer hardware can be found in [209]. Goniometer signals were acquired using an A/D card (NI PCI6071E, National Instruments, Austin, TX, USA) in the same xPC Target Computer responsible for control of the prosthesis (Section 4.4.4). Joint angular velocities were obtained using numerical differentiation and filtering in xPC Target. Two force-sensing resistors (FSRs) were attached to the shoe sole of the prosthetic leg to detect heel strike and toe off for data analysis purposes.

5.2.2 Commercial Prosthesis

For comparison, the subject also walked using his own prosthesis: an Otto Bock Genium (Otto Bock, Duderstadt, Germany, Figure 5.2). It presents the latest evolution of variable-damping devices. Compared to the C-Leg, its predecessor, it uses new sensors (gyroscope, accelerometer) in addition to angular sensor and strain gauge in the shank. Sensor data is acquired by the microprocessor at a sampling rate of 100 Hz, an increase of factor two compared to the previous generation. The actuation is similar to the old C-Leg, featuring a variable hydraulic damper. It weighs approximately 1.6 kg without a prosthetic foot. It has been well received by patients [66].
5.3 Experimental Conditions

The subject performed gait analysis sessions on two days, with one week in between. The first day, the prototype was adjusted to fit the amputee shaft and leg length, and a gait analysis using CLME control (Subsection 5.3.2) was performed. The subject selected a speed of 2.2 km/h on the treadmill, which was then also used for the following conditions. On the second day, the analysis was repeated using state-machine control (Subsection 5.3.1), and with his own prosthesis, the Otto Bock Genium knee commercial variable-damping prosthesis used for comparison.
5. Case Study

Genium knee (Subsection 5.3.3). In total, this yielded three different conditions that were compared (two with the developed prototype and one with the variable-damping device).

For each of the tested conditions, the analysis started with relaxed standing trials that were used as a reference for the joint angular displacements, followed by calibration trials where a specific joint was moved to determine functional joint centers [145].

5.3.1 State-Machine Control

After the calibration trials to determine functional joint centers for motion capture, the subject had time to familiarize himself with the prosthetic control on even ground with hand rails. He chose to change to the treadmill after about 5 minutes, in order to perform more consecutive steps without changing directions. The subject walked on the treadmill for about 10 minutes with short breaks in between to tune control parameters. Data were collected towards the end of those 10 minutes.

A finite-state controller was employed as commonly done in active transfemoral prostheses [18, 70, 74], in which a gait cycle was divided into four states associated with different impedance parameters. The control law in each state can be described as

$$\tau_{\text{ref}} = K (\phi - \phi_{\text{ref}}) - B \dot{\phi},$$

(5.1)

where $\tau_{\text{ref}}$ represents the reference knee moment commanded by the controller, $K$ represents virtual stiffness with setpoint (or equilibrium angle) $\phi_{\text{ref}}$, and $B$ virtual damping. These three parameters, as well as the switching conditions between the four states, were tuned manually. The final parameters are reported in Table 5.1.

<table>
<thead>
<tr>
<th>State</th>
<th>$K$ (Nm/rad)</th>
<th>$\phi_{\text{ref}}$ (°)</th>
<th>$B$ (Nms/rad)</th>
<th>transition condition</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>150</td>
<td>8</td>
<td>1</td>
<td>$\phi &gt; 10^\circ$ &amp; $\dot{\phi} &gt; 5.7^\circ$/s</td>
</tr>
<tr>
<td>2</td>
<td>80</td>
<td>7</td>
<td>0</td>
<td>$\phi &gt; 20^\circ$ &amp; $\dot{\phi} &gt; 5.7^\circ$/s</td>
</tr>
<tr>
<td>3</td>
<td>50</td>
<td>70</td>
<td>0</td>
<td>$\phi &gt; 50^\circ$ &amp; $\dot{\phi} &lt; 0^\circ$/s</td>
</tr>
<tr>
<td>4</td>
<td>0</td>
<td>20</td>
<td>1</td>
<td>$\phi &lt; 15^\circ$</td>
</tr>
</tbody>
</table>

Table 5.1: Parameters in state-machine control. When the indicated transition condition is fulfilled, the controller switches to the next state in the table (except from 4 it switches back to 1).

5.3.2 CLME Control

After the calibration trials to determine functional joint centers for motion capture, a further standing trial and a sitting trial were necessary to calibrate gain and offsets of the goniometer sensors. After that, the subject had time to familiarize himself with the prosthetic control on even ground with hand rails. He chose to change to the
treadmill after about 5 minutes, in order to perform more consecutive steps without changing directions. He walked for about 10 minutes, during which he experimented with how the prosthesis behaved in response to actions of his sound limb. Kinematics and prosthesis control signals were collected towards the end of these 10 minutes.

CLME control was based on previous work by Vallery et al., where prosthetic knee angle was predicted by from residual body motion, i.e. from angular data from sound side hip and knee using linear regression [164]. This concept was extended to also predict prosthetic knee stiffness $K$, and the prosthesis was controlled by an impedance control scheme

$$\tau_{\text{ref}} = K (\phi - \phi_{\text{ref}}) - B \dot{\phi}, \quad (5.2)$$

similar to the control law described above, but $K$ and $\phi_{\text{ref}}$ are now a continuous functions of residual body motion. The control scheme is described in detail in Appendix C. It involves linear regression to estimate $K$ and $\phi_{\text{ref}}$ from angular data from the sound hip and knee angle and angular velocities measured using goniometers (Subsection 5.2.1). During familiarization with the control law the subject reported that the behavior was too stiff, and stiffness was scaled by a constant factor of 0.5.

### 5.3.3 Control Scheme in the Commercial Prosthesis

Using his own prosthesis, the subject directly went on the treadmill after motion capture calibration trials. Gait data was collected after about 3 minutes of treadmill walking.

The control law employed in the subject’s Genium knee (Otto Bock, Duderstadt, Germany) is based on a state-machine. It adapts the damper in the prosthesis depending on estimated gait phase, i.e. a high damping in stance phase, and a low damping in swing phase. While internal details of the control strategy are unknown, some details have been published. In walking mode, which was used in this study, the knee maintains a knee angle of 4° prior to heel strike. Knee stance flexion (also called yielding) reaches up to 17° knee flexion, depending on the situation. The prosthesis adapts to walking speed and step length, though the algorithms how this is done are unknown. Furthermore, using a movement similar to backwards-kicking, obstacles can be stepped over using the prosthesis. It also features many different modes such as cycling, skiing or similar, which the user can choose using a remote control or by tapping on the floor with the prosthesis. In the present study only level-ground walking was investigated.

### 5.4 Data Acquisition

#### 5.4.1 Motion Capture and Marker Setup

Kinematics were captured using a 10-camera Vicon optical tracking system (Vicon Motion Systems Ltd., Oxford, United Kingdom). Marker data was sampled at
200 Hz. The marker setup closely followed the setup described by Wolf et al. [145], except for the feet, which used less markers because angles of the metatarsophalangeal joint were not evaluated. The reduced markerset is depicted in Figure 5.3. Joint angles were obtained using analysis described in [145].

5.4.2 Data Recording from the Prosthesis Setup

Data acquisition methods for sensor signals from the prosthesis are described in Section 4.4.3. The FSRs were connected to the STM32 microcontroller on the prosthesis, and their signals were sent to the xPC Target using the same RS-485 bus as the rest of the prosthesis sensor signals.

To synchronize measurements of the Vicon system with the signals acquired through xPC Target, a trigger was used to start both acquisition systems at the same time. Slight differences in acquisition clocks of both systems resulted in the measurements drifting slowly apart in both systems. This was corrected for by synchronizing both data sets using a signal that was present in both systems: the knee angle of the prosthesis. The signals were resampled such that the maximum correlation between the prosthetic knee angle recorded by the different systems was achieved.

5.5 Data Analysis

Kinematics of knee, hip, and ankle joint of sound limb and prosthetic limb were analyzed. The coordinate systems of the hip, knee and ankle angles were functionally
defined, and joint angles were calculated using a helical axis approach, both as described by Wolf et al. [145]. Five consecutive gait cycles were analyzed. The cycles were segmented based on the heel strike as detected by the force sensing resistor, and cycle time was normalized to a full cycle from heel-strike to heel-strike of the same leg.

For CLME and state-machine control, further information on behavior of the prosthesis was available, in particular the knee moments as measured by the observer (Section 4.4.3), as well as reference stiffness and equilibrium angle. The knee moments were compared between the two conditions. Stiffness and equilibrium angle were also compared to a controller reported in the literature [70]. It should be noted that the profiles from [70] were used in treadmill walking at a speed of 5.1 km/h, and that the exact state transition behavior was unknown, it was only reported in which periods of the gait cycle (in %) those values were active. Moment tracking performance, which allows to assess performance of the low-level control of the actuator, was also determined from this experiment. Results were presented in Chapter 4, Section 4.5.

Kinematics, knee moments, and stiffness estimates were also compared to data from an able-bodied subject (age 29, height 1.84m, weight 72kg) walking on even ground at a speed of 3.5 km/h. These data were already used in Section 3.4. This data was preferred over normative data from the literature because the same experimental conditions and processing methods could be used, making them more comparable.

5.6 Results

5.6.1 Gait Kinematics

Both control strategies using the powered prototype showed stance flexion (also called yielding) after heel-strike (Figure 5.4, right). The variable-damping knee yielded less. The prototype and the variable-damping knee reached full extension in stance phase sooner than the able-bodied subject, hence, not fully replicating physiological stance kinematics. State-machine control led to a higher peak flexion in swing phase than all other experimental conditions or the able-bodied pattern.

The knee kinematics of the sound limb were similar in all cases, but state-machine and variable-damping knee resulted in a slight flexion offset of approximately 5° compared to able-bodied pattern and CLME control, leading to hyperextension of approximately 5° (Figure 5.4, left).

Hip kinematics for the sound leg were very similar using state-machine control and the variable-damping knee, very close to an able-bodied pattern (Figure 5.5, left). With CLME control, an offset of approximately 10-15° could be observed before and after heel strike.
5. Case Study

On the amputated side, hip kinematics using the powered prototype slightly differed from the able-bodied pattern at the end of swing phase using both state-machine and CLME control: At the end of the gait cycle, a higher hip flexion could be observed (Figure 5.5, right). In the variable-damping device, this effect was visible but less pronounced.

Ankle kinematics were similar in all cases, but the sound side and the prosthetic side differed, due to the passive foot. The range of motion on the prosthetic (passive) ankle was much smaller than in a physiological ankle (Figure 5.6). After heel-strike, the prosthetic plantarflexion angle was around 8° lower than on the sound side. At toe-off, no plantarflexion could be observed in the prosthetic ankle, while on the sound side and in physiological gait it was over 15°.

5.6.2 Knee Kinetics

Compared to able-bodied knee moments determined using inverse dynamics, the powered prototype measured lower peak flexion and extension moments (Figure 5.7). Both control schemes, state machine and CLME, produced very similar prosthetic knee moments, a small difference could only be observed in swing phase. For the commercial variable-damping knee, no moment measurements were available, and the treadmill did not include force plates that would have been necessary for inverse dynamics.
5.6. Results

Figure 5.5: Hip kinematics with different control strategies for the unimpaired leg (left) and prosthetic leg (right) in treadmill walking. Mean and standard deviation over five gait cycles are shown.

Figure 5.6: Ankle kinematics with different control strategies for the unimpaired leg (left) and prosthetic leg (right) in treadmill walking. Mean and standard deviation over five gait cycles are shown.

5.6.3 Prosthetic Knee Stiffness and Setpoint

Stiffness values as commanded by the two control schemes (state machine and CLME) were compared to values estimated in physiological gait (see Section 3.4). Physiological setpoint was estimated as described in Appendix C. To relate these values to the literature, the comparison also included values reported by Sup et al. [70]
5. Case Study

Figure 5.7: Knee moment for the two control schemes used in the powered prototype compared to an able-bodied gait pattern. Mean and standard deviation over five gait cycles are shown.

that were used to control their powered transfemoral prosthesis. Stiffness estimated in physiological gait is much higher than stiffness of all the control schemes throughout most of the gait cycle (Figure 5.8). In particular, peak stiffness during stance is 440 Nm/rad for the physiological estimate, 150 Nm/rad for our state-machine control, 230 Nm/rad for state-machine control by Sup et al., and 106 Nm/rad for CLME control. During swing phase, physiological stiffness reached a minimum of 63 Nm/rad, while stiffness in our state machine drops commanded stiffness to zero, and Sup’s statemachine to 6 Nm/rad. The shape of the stiffness profile commanded by CLME control resembles physiological stiffness, while the shape of the statemachines’ stiffness was more abstract. It should be noted that the commanded stiffness for CLME control illustrated here was not actually achieved during the whole gait cycle: Shortly before toe off, moment tracking performance was insufficient due to motor saturation (see Figure 4.15 in the previous chapter).

The setpoint profiles in all cases are more or less constant around 5-10 ° throughout most of the stance phase, and increase to a peak of 42 ° using CLME control, 70 ° for our state-machine control, and 65 ° for Sup’s controller (Figure 5.8). Estimated physiological setpoint peak is 51 °. The setpoint in CLME control starts to increase already at 45% gait cycle, while for the other control schemes it increases later, at around 55-65% in the cycle.
5.7 Discussion

The goal was to test in a case study whether the actuator designed based on requirements for physiological capabilities could restore physiological gait patterns when used in a transfemoral prosthesis. An experiment was performed that involved treadmill walking using two control strategies with the ANGELAA-leg (the powered prototype), in which near-natural gait kinematics could be achieved. Gait kinematics with the ANGELAA-leg were also compared to kinematics acquired using a state-of-the-art variable-damping device, which were very similar.

5.7.1 Kinematics of the Prosthesis and Physiological Gait

Overall, the kinematics of all joints using the three experimental conditions were fairly similar, and similar to physiological gait data reported in the literature [121, 145, 210]. Quantitative comparisons to data from the literature are difficult due to
different measurement protocols and different definitions of the zero-positions of the joints. Within a single study, inter-subject variability was reported to be around 7-9° SD averaged over cycle time [121].

Prosthetic knee angle was surprisingly similar for both control strategies with the powered prototype ANGELAA and the variable-damping device. Yielding at the beginning of stance phase was smaller in the variable-damping knee, likely due to the limitation that energy can only be dissipated. Peak knee flexion angle with state-machine control was higher than for all other experimental conditions or the healthy leg, and could probably be reduced by more extensive tuning of the control parameters.

More differences could be observed in the sound limb. Using CLME control, knee angles during the first 30% of the gait cycle were about 10° higher than with state machine or variable-damping control. This may be explained by adaption of the sound leg behavior to effect the desired motion in the prosthesis, which is controlled through sound leg motion in CLME control. This is supported by feedback of the subject, who reported that he had to get used to the control scheme, because it acted fundamentally different than his own prosthesis, forcing him to focus on movement of the sound limb, rather than on behavior of the prosthesis. It should be noted that in the amputee subject, sound leg knee angles reached hyperextension of up to 6°. This is likely a calibration artifact, and it at least partially explains the decrease of knee flexion in the sound leg. All joint angles were calculated in relation to relaxed standing (in accordance with [145]), and the amputee probably stood with his healthy knee slightly flexed, to have a better balance. To correct for this in retrospect, healthy knee angles should be offset by about 5°. This would bring the profiles using state-machine and variable-damping control very close to the physiological pattern, and it would offset the profile using CLME control to slightly higher flexion angles than in the physiological pattern. Further experiments should thoroughly exclude such calibration artifacts.

Hip angle of the sound leg was very close to physiological gait using both state-machine control and the variable-damping knee. Using CLME control, there was an offset of about 10°, probably also due to adaption to effect desired prosthesis motion.

Hip angle of the prosthetic limb was similar in all experimental conditions, and close to physiology, except at the end of swing phase, where hip flexion was higher using the ANGELAA-leg, with both control schemes, compared to the variable-damping knee or the physiological pattern. A possible explanation is the following: using CLME control, prosthetic knee extension at the end of swing phase substantially slowed down already around 15° flexion (Figure. 5.4), and using state-machine control, extension occurred later than with the variable-damping knee the subject was used to, and later than in physiological gait. Therefore, it seems likely that this larger hip flexion represents an effort by the amputee to completely extend the
prosthetic knee before heel strike. For purely mechanical prostheses it is important that the knee is completely extended upon heel strike, as they collapse otherwise; the Genium knee that the subject normally wears does not have this limitation. Nevertheless, it is possible that the subject aimed to increase stability by trying to extend the prosthetic knee as far as possible using the hip joint.

5.7.2 Stiffness of the Prosthesis and Physiological Stiffness

For a function that was perceived comfortable by the subject, the knee stiffness rendered using the prosthesis was much lower than what the model-based approach to estimate short-range stiffness predicted for a physiological knee. This observation could be made for both control strategies. For state-machine control, the stiffness parameters were tuned manually without considering physiological estimates, and peak values were about factor three below physiological estimates. For CLME control, commanded stiffness in Equation 5.2 was manually reduced by a factor of two. Another element resulted in an even further decrease of stiffness: stiffness predicted by CLME control (with the mapping parameters from Appendix C) was already lower than physiological stiffness, so the resulting peak values (including the manual reduction by a factor two) were about factor four lower than physiological estimates. This lower stiffness in the CLME predictions may be due to a different gait pattern in training data used for the CLME control design (see Appendix C) when compared with the pattern from the amputee. The lower CLME prediction may also be in part attributed to the velocity-dependent mapping of stiffness (see parameter $\alpha$ in Appendix C).

The reduced stiffness in the prosthesis in either control strategy compared to physiological stiffness estimates could have several explanations. To some extent, a slightly lower required stiffness could be expected due to the lower walking speed of the amputee (2.2 km/h) compared to the able-bodied subject (3.5 km/h), because knee moments increase with increasing walking speed [211], which in turn increases stiffness. This could in part also explain the lower knee moments observed in the ANGELAA-leg compared to physiological gait. However, stiffness was reduced more than what could be expected from the difference in gait speed, and also a few tests with the prosthesis at higher speeds did not require higher stiffness of the control strategy.

A more probable explanation is that model-based stiffness estimates could be too high. As discussed in Chapter 3, the estimation approach has only been validated in isometric conditions, and it is possible that it cannot be directly transferred in this form to movements. There are indications in the literature that joint stiffness may be lower during movement than what would be expected from observations in the isometric conditions [212]. In particular, in perturbation experiments a reduction in stiffness was observed during movement onset [39,159].

Another explanation could be that short-range stiffness is not a good predictor of
joint behavior during gait. The experiments in Chapter 2, Section 2.4.3 showed that measured knee joint stiffness decreases with perturbation amplitude, as has been shown in the literature [41, 106]. It is possible that stiffness in response to larger excursions better reflects joint behavior during gait, and other contributors to joint behavior have to be considered in more detail. Human motor control is commonly thought of consisting of several layers that are tightly interlinked: cerebral control, spinal reflex loops, and intrinsic muscle stiffness [213]. Cerebral control may possibly be replaced by the intention estimation of a prosthesis. However, additional reflexive components to stiffness may have to be considered when physiological stiffness estimates are used to mimic physiological behavior in a prosthesis.

It is also likely that physiological stiffness, even if it could be determined perfectly, should not directly be copied in a prosthetic device. Controlling a prosthesis is something very different from controlling a physiological knee that is tightly integrated into the human sensory-motor loop. First, the mechanical attachment differs greatly, and stiffness might be perceived as comfortable when for example movement of the shaft with respect to the residual limb is minimal, and not when stiffness replicates physiological stiffness. There is also only limited sensory feedback from the prosthetic leg and actuator, and a movement of a prosthetic joint might be perceived stiffer than the movement of a physiological joint, where constant afferent feedback is available.

Furthermore, inertial properties of the prosthesis are different than inertial properties of a physiological shank. Inertia of the prosthetic shank is about factor two lower, and natural swing frequency is about 15% higher. While the influence of these differences should be low at least during stance phase due to the small accelerations, it likely requires stiffness to change in relation to inertial properties in swing phase.

It should also be noted that using CLME control, the motor torque reached saturation near toe-off around 55-62% gait cycle, as discussed in Chapter 4, Subsection 4.5.6. Therefore, stiffness depicted in Figure 5.8 could not be rendered during this period for CLME control.

Interestingly, knee moments and gait kinematics were very similar using both control approaches, even though stiffness profiles were different. This suggests that stiffness might not be the most important factor to replicate physiological gait patterns, at least not during the whole gait cycle in level-ground walking. Nevertheless, it should be highlighted that the subject felt uncomfortable when a higher stiffness was used, indicating that stiffness is an important facet in control of powered prostheses.

5.7.3 Further Limitations of the Study

One limitation of the study is that the device was only tested with one amputee subject. However, the focus was to investigate the performance of using the presented actuator principles in a transfemoral prosthesis. Also the physiological gait data that
5.7. Discussion

were used for evaluation and extraction of the CLME mapping were based on one subject only. Adding more subjects likely would have increased variability in the kinematic data. As kinetic, kinematic and EMG data were consistent with the literature [47,121,210], this is not considered a big shortcoming of the study.

The proposed control schemes allowed the replication of near-natural gait kinematics, but only limited time was spent for tuning control parameters, and more extensive tuning may further improve gait characteristics. Furthermore, a method for automatic parameter tuning in state-machine control, or a method to find initial estimates, may shorten tuning time [75,76].

A passive foot was used, which inherently limits reproduction of physiological gait, at least kinematics of the prosthetic ankle joint, compared to devices including an actuated ankle [71]. Nevertheless, near-physiological gait kinematics were achieved in the other joints of both legs.

It should also be noted that the amputee walked at a self-selected speed of 2.2 km/h, which is relatively slow, and it would be interesting to investigate higher speeds in the future.

Finally, the advantages of a powered device are most prominent in tasks where much kinetic energy generation is required, such as stair climbing [121] or sit-to-stand transfer [13]. In those tasks, stiffness potentially plays a bigger role, and accurate replication of stiffness may become more important.

5.7.4 Conclusion

The proposed actuator prototype ANGELAA could be used to restore almost natural gait kinematics in a transfemoral amputee with very little familiarization time. Such near-physiological gait kinematics could be achieved using substantially different profiles for knee stiffness. Nevertheless, the amputee perceived stiffness profiles that were lower than physiological estimates as more comfortable, which suggests that stiffness is an important factor that needs to be considered in prosthesis design and control. The lower stiffness profiles could also indicate that so far not modeled neural commands such as reflexes might influence joint impedance substantially and should be considered in future modeling and control approaches.
Conclusion & Outlook

6.1 Conclusion

Thanks to advances in energy-efficient actuators and energy storage technology, powered prostheses have become a possibility today. While this gives hope to amputees that they will be able to perform similar activities as able-bodied people, existing artificial limbs are still limited compared to their physiological counterparts. The aim of this thesis was to contribute towards the design and control of transfemoral prostheses that ultimately improve quality of life of amputees. The underlying assumption was that physiological joint stiffness caused by short-range stiffness of muscle should be considered for the design and control of transfemoral prostheses. Specifically, the goals of this thesis were:

- Development of a method to quantify physiological knee stiffness
- Derivation of requirements for design and control of transfemoral prostheses
- Design of an actuation principle based on these requirements
- Evaluation of the performance of the actuation principle in a transfemoral prosthesis

These goals were achieved as follows:

A model-based approach to quantify physiological knee stiffness was developed. It does not require the application of perturbations to the joint, which was necessary in approaches to estimate knee stiffness from the literature. The approach was validated by comparison to perturbation measurements in isometric conditions including conditions with significant co-contraction. Previously, knee stiffness has only been investigated without controlled amounts of co-contraction, even though it is known that co-contraction plays an important role in human motor control. The presented
approach relies on EMG signals, kinetic and kinematic measurements. It can predict knee joint stiffness in conditions where these signals can be measured. This presents an important step in the process of understanding and replicating physiological motor control in a variety of activities, where EMGs, kinetics and kinematics are measured. The approach to estimate knee stiffness uses scalable muscle models and presents a general approach that could also be transferred to other joints throughout the body.

Commonly, requirements for transfemoral prostheses are derived in terms of moment generation and velocity, and not in terms of joint stiffness. In this thesis, a detailed derivation of joint stiffness requirements for transfemoral prostheses was presented. This was possible by leveraging the model-based approach to quantify stiffness. All requirements were based on the peak capabilities of a physiological knee. In particular, the peak joint stiffness as a function of angle was analyzed, to derive stiffness requirements for the elastic element in series-elastic actuators (SEAs). To the author’s knowledge, this presents the first approach to estimate the stiffness needed to replicate physiological joint stiffness in a prosthesis. Other requirements that were analyzed include moment generation in function of joint angle and velocity, joint power, joint velocity and mechanical eigenfrequency of the lower leg. All requirements can be used as a reference for future designs of prostheses and other devices where peak physiological knee joint capabilities play an important role, such as orthoses or training devices.

Based on these requirements, an actuation concept was designed and realized. Compared to other transfemoral prostheses, which usually employ regular SEAs or no elasticity at all, the developed actuation concept employs a novel variant of series-(visco)elastic actuator, that changes its physical stiffness in function of kinematic configuration, optimized for a physiological stiffness profile. The concept contains two special measures to reduce weight: rubber cords as series-elastic elements, and nonlinear parallel elasticity to reduce motor torque requirements. The parallel elasticity creates equilibria near $0^\circ$ as well as $90^\circ$ knee flexion, frequent postures in daily life. The principles employed in the actuator can also be applied to other impedance-controlled devices where stiffness requirements can be specified in advance and formulated in function of kinematic configuration. Examples are other wearable devices like ankle prostheses or leg exoskeletons, but also mobile manipulators.

Finally, the actuator was tested in a transfemoral prosthesis prototype with an amputee subject using different control strategies. The device was named ANGELAA (ANGle-dependent ELAstic Actuator). One control strategy was based on a finite state machine that switched impedance parameters based on estimated gait phase, similar to most approaches presented in the literature. The other strategy inferred prosthesis stiffness and equilibrium angle continuously from motion of the sound leg. It presents an extension of an existing control strategy that worked in
6.2. Outlook

Future work may address several areas. The model-based method allows accurate quantification of joint stiffness in static conditions. Evaluation in dynamic conditions, in particular during gait, would be interesting, but experimental challenges with performing such measurements need to be solved first. An actuated orthosis recently developed for this purpose [45] may be used to tackle these challenges. To extend the model, viscous properties of the musculoskeletal system may also be considered in the future.

The prototypical hardware is designed to enable rendering of different stiffness profiles in the same range as a physiological knee, such that it can be used as a research platform for different control strategies. For a long-term evaluation, control strategies need to be tested outside the laboratory in real-life situations, and, hence, a self-contained device should be developed. For this purpose, energy efficiency is crucial, and it may be beneficial to employ actuators that can arbitrarily vary their stiffness mechanically (e.g. [18,191]), rather than through control.

The control strategies used to evaluate the developed device were able to reproduce natural kinematics in treadmill walking. To develop a control strategy that truly mimics biology with all its different facets and layers, several components would likely have to be added: First, it is known that afferent feedback loops (also called reflexes) play a crucial role in control of human gait [27,28], and, therefore, should be integrated into the control scheme, as already tested in ankle prostheses [214,215]. Second, in addition to or alternative to the employed CLME, further ways of conveying voluntary commands to the prosthesis could be explored. One way would be the use of surface EMG signals [72,78], in particular in combination with targeted muscle reinnervation (TMR) [83]. Third, also sensory feedback from the leg to the human should be restored, to better integrate the prosthesis in the sensory-motor

stiff position control mode using the same principle and is called complementary limb motion estimation (CLME). The ANGELAA-leg was able to reproduce near-natural gait kinematics in treadmill walking with both control strategies, even though prosthetic stiffness profiles were different in both approaches. The case study also included comparison to a state-of-the-art variable-damping prosthesis, which allowed similar gait kinematics.

While the surface of complete understanding of the relevant physiological principles has only been scratched, this thesis presents a piece in the long process of understanding human motor control for biomimetic control of prostheses, and it presents examples of how biomimetic principles can be exploited to design appropriate actuators. The technological concepts may contribute to develop cheaper, lighter, and more intuitive prosthetic products, that make intelligent prostheses available to a larger amputee population and easier to use, helping to relieve their handicap.
control loop, and to facilitate coordination between physiological and artificial limbs. This could be done for example using sensory substitution techniques [216, 217], or by direct nerve stimulation [218].

Furthermore, mechanical fixation of leg prostheses to the stump remains one of the major challenges, which was not addressed in this thesis. Fixation with a shaft that covers the residual limb still allows the fixation to move with respect to the femur, and likely presents different requirements to the control of a prosthesis than if it was rigidly connected. Osseointegration may be a solution to this problem [219], if the main problems associated with it, such as higher risk of infections and longer rehabilitation process [220], can be addressed.

Finally, it should be noted that this work was based on the assumption that a prosthetic leg should replicate a physiological leg. While this is certainly a wish of many amputees, it is possible that an artificial leg could be more useful if it did not strictly follow physiological examples, and future work could explore such possibilities.
Physiological Background on Muscle Stiffness

To understand the functional properties of skeletal muscle and muscle stiffness, basic knowledge of muscle structure is important. A muscle is commonly connected to a bone through a tendon. The muscle consists of many muscle fibers, which contain multiple myofibrils in parallel, and those contain many sarcomeres in series. These basic structures are depicted in Figure A.1. The sarcomeres are the force-generating elements in muscle and are around 1-3 micrometers in length [221]. The force is generated through interaction of actin and myosin filaments. The number of sarcomeres in series in myofibrils are about 30,000-40,000 in human quadriceps muscles [222].

![Figure A.1: Structural elements of a skeletal muscle (figure adapted from [223]).](image)

A widely accepted theory of how sarcomere force is generated is through sliding of actin and myosin filaments past each other [224, 225]. Huxley proposed that the sliding of the filaments is caused by forming of cross-bridges between actin and myosin filaments and that these cross-bridges contain an elastic element [226] (Figure A.2). Gordon et al. [227] found that the size of the overlapping region of actin and myosin filaments can explain the well-known force-length relationship of muscle first discovered by Hill [228]. The sliding-filament theory can also explain the
A. PHYSIOLOGICAL BACKGROUND ON MUSCLE STIFFNESS

force-velocity relationship of sarcomeres, which is largely determined by the rate of formation and breakdown of cross-bridges, which happens in a cyclic manner [226]. While the basic aspects of the cross-bridge theory are commonly accepted, the exact mechanism causing the filaments to slide is still under debate [229–231].

Figure A.2: Muscle contraction is caused by sliding filaments inside the sarcomere. The sliding of the filaments is caused by formation of cross-bridges between thick (myosin) and thin (actin) filaments (figure adapted from [232]).

Muscle forces in response to rapid perturbations with a small amplitude are governed by the short-range stiffness, which has been attributed to the elasticity of the cross-bridges [14]. It has been shown that short-range stiffness largely depends on muscle force, and, hence, the number of attached cross-bridges [98]. Other mechanisms may also contribute to short-range stiffness, such as the compliance of the titin filaments [233]. Their properties are mainly relevant for properties of relaxed muscle [233], but theories exist that cross-bridge cycling causes the titin filament to be wound around the thin filament, thereby influencing sarcomere stiffness in activated muscle [231]. The short-range stiffness must not be confused with the slope of the force-length relationship of muscle [24,234]. Even though the physiological mechanisms behind short-range stiffness are not entirely understood, functional models that relate muscle force to short-range stiffness have been successfully developed [100].
Actuator Geometric Mappings

In this appendix, the geometric relationships in the series-elastic mechanism in the prosthesis are derived. The variables used here are shown in Figure B.1. Constants that are known from the design are: \( a, b, c_1, c_2, \phi_0, \delta_0, \theta_1, r_1, r_2, l_{01}, l_{02}, K_1, K_2 \). Here, linear purely elastic springs with stiffness \( K_1, K_2 \) are assumed, in contrast to the knee moment observer in Section 4.4.3. The lengths \( x, l_a, l_1, l_2 \) vary with rocker angle \( \alpha \) and knee angle \( \phi_p \). They are given by the following equations:

\[
x^2 = a^2 + b^2 - 2ab \cos (\phi_0 + \phi_p) \tag{B.1}
\]

\[
\epsilon = \arcsin \left( \frac{a}{x} \sin (\phi_0 + \phi_p) \right) \tag{B.2}
\]

\[
l_a^2 = r_2^2 + x^2 - 2r_2x \cos (\delta_0 - \alpha - \epsilon) \tag{B.3}
\]

\[
\cos (\gamma) = \frac{r_2^2 + l_a^2 - x^2}{2r_2l_a} \tag{B.4}
\]

\[
F_a = \frac{F_a \perp}{\cos (\gamma)} \tag{B.5}
\]

\[
l_1^2 = c_1^2 + r_1^2 - 2c_1r_1 \cos (\theta_1 + \alpha) \tag{B.6}
\]

\[
l_2^2 = c_2^2 + r_1^2 - 2c_2r_1 \cos (\theta_1 - \alpha) \tag{B.7}
\]

Assuming linear elastic springs, the forces in the springs \( F_1, F_2 \) and the resulting actuator force \( F_a \) and knee moment \( \tau_a \) are:

\[
F_1 = K_1 (l_1 - l_{01}) \tag{B.9}
\]

\[
F_2 = K_2 (l_2 - l_{02}) \tag{B.10}
\]

\[
\zeta_1 = \arccos \left( \frac{l_1^2 + r_1^2 - c_1^2}{2l_1r_1} \right) \tag{B.11}
\]

\[
\zeta_2 = \arccos \left( \frac{l_2^2 + r_1^2 - c_2^2}{2l_2r_1} \right) \tag{B.12}
\]
**B. Actuator Geometric Mappings**

![Prosthesis sketch and variables.](image)

Figure B.1: Prosthesis sketch and variables.

\begin{align*}
F_{1\perp} &= F_1 \sin(\zeta_1) & \text{(B.13)} \\
F_{2\perp} &= F_2 \sin(\zeta_2) & \text{(B.14)} \\
F_a \perp &= \frac{r_1}{r_2} (F_{1\perp} - F_{2\perp}) & \text{(B.15)} \\
F_a &= \frac{F_a \perp}{\cos(\gamma - \pi/2)} & \text{(B.16)} \\
\tau_a &= r_a F_a & \text{(B.17)}
\end{align*}
The actuator moment arm can be calculated using

\[
\beta = 2\pi - (\phi_0 + \phi_p) - (\delta_0 - \alpha) - \gamma \\
y^2 = a^2 + l_a^2 - 2al_a \cos(\beta) \\
\eta = \arccos \left( \frac{y^2 + l_a^2 - a^2}{2yl_a} \right) \\
r_a = y \sin(\eta)
\] (B.18, B.19, B.20, B.21)

As a function of \(\phi_p\) and \(\alpha\), the knee moment can be written as

\[
\tau_a = r_a F_a \\
= r_a \frac{F_a}{\cos(\gamma - \pi/2)} \\
= r_a \frac{F_a}{\sin(\gamma)} \\
= r_a \frac{1}{\sin(\gamma)} \frac{r_1}{r_2} (F_{2,\perp} - F_{1,\perp}) \\
= R (F_{1,\perp} - F_{2,\perp}) \\
= R (F_1 \sin(\zeta_1) - F_2 \sin(\zeta_2)) \\
= R (K_1 (l_1 - l_{01}) \sin(\zeta_1) - K_2 (l_2 - l_{02}) \sin(\zeta_2)) \\
= R (K_1 (l_1 - l_{01}) S_1 - K_2 (l_2 - l_{02}) S_2) \\
= RS_1 l_1 K_1 - RS_1 (K_1 l_{01}) - RS_2 l_2 K_2 + RS_2 (K_2 l_{02}) \\
= (RS_1 l_1, -RS_2 l_2, -RS_1, RS_2) \cdot (K_1, K_2, K_1 l_{01}, K_2 l_{02})^T
\] (B.22-31)

where

\[
R = r_a \frac{1}{\sin(\gamma)} \frac{r_1}{r_2} \\
S_1 = \sin(\zeta_1) \\
S_2 = \sin(\zeta_2)
\] (B.32-34)
APPENDIX

CLME Control Details

The CLME control law implemented in the prototype is given by

$$\tau_{\text{ref}} = \dot{K} (\phi - \hat{\phi}_{\text{ref}}) - B \dot{\phi},$$

(C.1)

where $\tau_{\text{ref}}$ is the reference moment sent to the prosthesis, $\dot{K}$ the commanded knee stiffness, $\phi$ the actual knee angle, $\hat{\phi}_{\text{ref}}$ the equilibrium angle, and $B$ the damping. The damping $B$ was set to a constant of 0.5 Nms/rad. Knee stiffness $\dot{K}$ and equilibrium angle $\hat{\phi}_{\text{ref}}$ are continuous functions of residual body motion as measured by the goniometers (Chapter 5, Subsection 5.2.1). These control variables $\hat{\phi}_{\text{ref}}$ and $\dot{K}$ are calculated using linear regression and a Kalman filter.

The control variables $\hat{K}$ and $\hat{\phi}_{\text{ref}}$ are calculated online, during wearing the prosthesis. First, stiffness $K$ and its derivative $\ddot{K}$ are calculated using

$$\left( \begin{array}{c} K \\ \dot{K} \end{array} \right) = \alpha \left( \begin{array}{c} \phi_h \\ \dot{\phi}_h \end{array} \right) + k_o,$$

where $\phi_h = (\phi_h, \dot{\phi}_h)^T$ are the joint angles of hip and knee of the sound leg. The values for $K$ and $\dot{K}$ are then fused using a Kalman filter to obtain a refined stiffness prediction $\dot{K}$ that is then used in (C.1).

The mapping parameters $\alpha$ and offset $k_o$ are determined offline in analogy to [164] from data collected of an able-bodied subject: The data included sensor signals of the goniometers (Chapter 5, Subsection 5.2.1), and physiological knee stiffness was estimated using the model-based estimation as described in Chapter 3, Section 3.4. The parameters $\alpha$ and $k_o$ are found using linear regression from these data, exactly as described in [164], but with different output signals. Here, joint angles and angular velocities of hip and knee of the sound leg $(\phi_h, \dot{\phi}_h, \dot{\phi}_k, \ddot{\phi}_k)^T$ are the input signals, and estimated stiffness $K$ and its derivative $\dot{K}$ the output signals. The obtained
values for the regression parameters were
\[
\alpha = \begin{pmatrix}
45.9 \text{Nm/rad}^2 & 71.0 \text{Nm/rad}^2 & -0.3 \text{Nms/rad}^2 & 20.4 \text{Nms/rad}^2 \\
-946 \text{Nm/(srad)}^2 & -1408 \text{Nm/(srad)}^2 & 123 \text{Nm/rad}^2 & 14 \text{Nm/rad}^2
\end{pmatrix},
\]

\[
k_o = \begin{pmatrix}
-29 \text{Nm/rad} \\
599 \text{Nm/(srad)}
\end{pmatrix}.
\]

The Kalman filter was implemented as in [164], but instead of angular signals the stiffness signals \(K, \dot{K}\) were used.

Using the same approach as for stiffness above, the equilibrium angle \(\phi_{\text{ref}}\) and its derivative \(\dot{\phi}_{\text{ref}}\) are defined as a continuous function of residual body motion and calculated online:
\[
\begin{pmatrix}
\phi_{\text{ref}} \\
\dot{\phi}_{\text{ref}}
\end{pmatrix} = \beta \begin{pmatrix}
\phi_h \\
\dot{\phi}_h
\end{pmatrix} + \phi_o.
\]

A Kalman filter is then used to merge the predictions of \(\phi_{\text{ref}}\) and \(\dot{\phi}_{\text{ref}}\) to obtain a refined control variable \(\hat{\phi}_{\text{ref}}\) in (C.1).

The mapping parameters \(\beta, \phi_o\) are calculated offline similarly as described above, and from the same data. First, the physiological knee moment \(\tau_{\text{phy}}\) is determined via inverse dynamics from recorded gait data of a healthy subject. Then, a trajectory for the equilibrium angle \(\phi_{\text{ref}}\) is determined that replicates this moment for the associated recorded physiological knee angles \(\phi_{\text{phy}}\), using the stiffness \(K_{\text{phy}}\), such that \(\tau_{\text{phy}} = -K_{\text{phy}} (\phi_{\text{phy}} - \phi_{\text{ref}})\). The parameters \(\beta, \phi_o\) are again found using linear regression as in [164], with the input signals \((\phi_h, \phi_k, \dot{\phi}_h, \dot{\phi}_k)^T\) and the output signals \(\phi_{\text{ref}}, \dot{\phi}_{\text{ref}}\). The resulting mapping parameters were:
\[
\beta = \begin{pmatrix}
0.24 & 0.07 & -0.15 \text{s} & 0.04 \text{s} \\
9.22 \text{s}^{-1} & 0.17 \text{s}^{-1} & -0.51 & 0.60
\end{pmatrix}, \quad \phi_o = \begin{pmatrix}
0.38 \text{rad} \\
-1.54 \text{rad/s}
\end{pmatrix}.
\]

The Kalman filter was implemented as described in [164].
Acronyms

CLME complementary limb motion estimation.
CoM center of mass.
EMG electromyogram.
FRF frequency response function.
FSR force-sensing resistor.
ICR instantaneous center of rotation.
MVC maximum voluntary contraction.
SD standard deviation.
SEA series elastic actuator.
TMR targeted muscle reinnervation.
VAF variance accounted for.
Glossary

$F_a$ Actuator pulling force (positive forces cause knee extension, negative forces cause knee flexion).

$F_a \perp$ Actuator pulling force perpendicular to rocker.

$l_a$ Actuator length from thigh pivot point to rocker pivot point.

$r_a$ Actuator moment arm with respect to knee rotation axis.

$\alpha$ Angle of the rocker that connects actuator to serial springs. Zero means no spring deflection, positive angle causes knee extension, negative causes knee flexion torque.

$f_M$ Muscle force.

$f_M^0$ Maximum isometric muscle force.

$f_T$ Tendon force.

$f_{MT}$ Musculotendinous force.

$\phi$ Knee flexion angle, full extension is defined as zero.

$\dot{\phi}$ Knee angular velocity.

$\tau$ Knee extension moment.

$l_M^0$ Optimal fiber length.

$l_{TS}^0$ Tendon slack length.

$\ddot{\phi}_m$ Motor acceleration.

$\varphi_m$ Motor position.

$\tau_m$ Motor torque.

$\dot{\varphi}_m$ Motor velocity.

$\alpha$ Pennation angle.
Glossary

$\phi_p$ Knee flexion angle of the prosthetic knee, full extension is defined as zero.

$\delta$ Rocker deflection angle with respect to the shank.

$k_M$ Muscle stiffness.

$k_T$ Tendon stiffness.

$k_{MT}$ Musculotendinous stiffness.
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Bibliography


List of Publications of Author

Journal Publications


Conference Publications


**Patent Publications**


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