FINITE STATE IMPEDANCE-BASED CONTROL OF A POWERED TRANSFEMORAL PROSTHESIS

By

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Dedicated to my mother, Moti and

my brothers, Hari and Sandeep

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CHAPTER I

MOTIVATION

Despite significant technological advances over the past decade, such as the introduction of microcomputer-modulated variable damping, commercial transfemoral prostheses remain limited to energetically passive devices. That is, the joints of the prostheses can either store or dissipate energy, but cannot provide net positive power over a gait cycle. The inability to deliver positive joint power significantly impairs the ability of these prostheses to restore many locomotive functions including walking up stairs and up slopes, running and jumping, all of which require significant net positive power at the knee joint, ankle joint or both [1-8]. Further, even normal walking requires positive power output at the knee joint and significantly net positive power at the ankle joint, (Fig 1-1) [1]. Transfemoral amputees walking with passive prostheses have been shown to expend up to 60% more metabolic energy relative to healthy subjects during level walking [8] and exert as much as three times the affected-side hip power and torque [1], presumably due to the absence of powered joints. Prosthesis with the capacity to deliver power at the knee and ankle joints would presumably address these deficiencies, and would additionally enable the restoration of normal biomechanical locomotion.

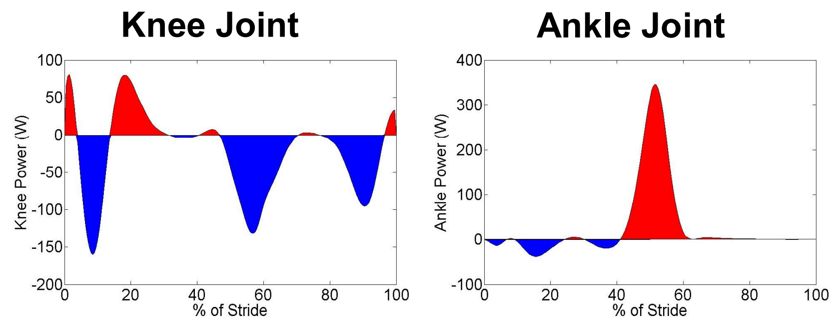


Figure 1-1. Joint power during one cycle for 75 kg normal subjects during fast walking. Red represents positive power generated the joints during level walk [1, 3].

A significant bottleneck to the development of powered prosthetics has been the lack of on-board power and actuation capabilities comparable to biological systems. State-of-the-art power supply and actuation technology such as battery/DC motor combinations suffer from low energy density of the power source (i.e., heavy batteries for a given amount of energy), low actuator force/torque density, and low actuator power density (i.e., heavy motor/gear head packages for a given amount of force or torque and power output). However, recent advances in power supply and actuation for self-powered robots, such as the liquid-fueled approaches described in [9-12] offer the potential of significantly improved energetic characteristics relative to battery/DC motor combinations and capable of biological scale energetic and power potential. This has brought the potential of powered lower limb prosthesis to the near horizon and led to the development of a prototype transfemoral prosthesis [41] that is intended to ultimately be powered by the liquid-fueled approach. The developed prototype is a fully powered two degree-of-freedom robot, capable of significant joint torque and power, which is rigidly attached to a user. As such, the prosthesis necessitates a reliable control framework for generating required joint torques while ensuring stable and coordinated interaction with the user and the environment. This paper proposes an impedance based control framework to produce normal gait patterns by characterizing joint behavior with a series of finite states consisting of spring and damper behavior. The impedance based characterization of knee and ankle joints allow for stable interaction while providing intuitive means of tuning the model to facilitate desired gait pattern. This allows for greater adaptability to different users in differing gait conditions.

CHAPTER II

PRIOR WORK

Prior work on transfemoral prosthesis can be understood in three broad categories, (a) Passive mechanical devices, (b) Microcomputer-based modulated damping devices and (c) Actively powered joints. Passive mechanical devices such as ‘peg’ legs or more sophisticated ‘four-bar’ or ‘six-bar’ linkage mechanics provide a purely mechanical framework for fulfilling simple necessitates of gait such as support and swing. However, since they do not contribute much to the control aspect, they are omitted here in discussion.

The initial ground work for the development of actively powered knee joints was the work by Flowers et al., which took place during the 1970’s and 1980’s by Flowers [13], Donath [14], Grimes et al. [16], Grimes [15], Stein [17], Stein and Flowers [18]. Specifically, Flowers and Mann [19] developed a tethered electrohydraulic transfemoral prosthesis that consisted of a hydraulically actuated knee joint tethered to a hydraulic power source and off-board electronics and computation. They subsequently developed an “echo control” scheme for gait control, as described by Grimes et al. [16], in which a modified knee trajectory from the sound leg is played back on the contralateral side. In addition to this prior work directed by Flowers, other groups have also investigated powered knee joints for transfemoral prostheses. Specifically, Popovic and Schwirtlich [20] report the development of a battery-powered active knee joint actuated by DC motors, together with a finite state knee controller [21] that utilizes a robust position tracking control algorithm for gait control [22]. They reported an increased cadence with lowering of metabolic energy

(O2 consumption) in clinical trials with amputees. However, the work was never commercialized due to limitations in power capabilities. With regard to powered ankle joints, Klute et al. [23-24] describe the design of an active ankle joint using pneumatic McKibben actuators, although gait control algorithms were not described. Au et al. [25] assessed the feasibility of an EMG based position control approach for a transtibial prosthesis.

Since power limitations have hindered the development of active joints, a large number of research has been devoted to the development of low power micro-computer controlled damping joints. Since the knee joint exhibits damping behavior for a significant part of level walk [1], this method does provide for feasible means of partial restoration of gait. Bar et al. [26] reported on the development of a microcomputer controlled knee joint. The gait was divided into 8 phases wherein damping at the knee joint was varied to enable successful gait using information from sound side. Goldfarb [27] described the development and control of knee prosthesis with modulated damping with no dependencies on the sound side. Herr and Wilkenfeld [28] further describe a user adaptive device with an autonomous algorithm that continuously adjusts the damping at the knee joint to enable gait at differing speeds. Since all of these devices are passive impedance-based, the control is limited to varying the damping values in different parts of the gait stride. In addition, Kim and Oh [29] also report a MR based legs and related tracking control.

Comparatively, since the ankle joint exhibits more of a spring like behavior during walk, prosthetic ankle joints with spring return have become increasingly popular. The stiffness is incorporated either using a mechanical spring or compliant material as in products offered by companies such as Ossur. However, the ankle is a very active joint producing impulsive power to aid the gait, and the spring behavior though helpful does not replicate ankle joint behavior. With springs incorporated, issues of control are minimal since no modulation of spring stiffness can be performed.

Finally, though no published literature exists, Ossur, a major prosthetics company based in Iceland, has announced the development of both a powered knee and a self-adjusting ankle. The latter, called the “Proprio Foot,” is not a true powered ankle, since it does not contribute power to gait, but rather is used to quasistatically adjust the angle of the ankle to better accommodate sitting and slopes. The powered knee, called the “Power Knee,” utilizes an echo control approach similar to the one described by Grimes et al. [16]

CHAPTER III

NORMAL WALK

Walking is a highly coordinated behavior accomplished by intricate interaction of the musculo-skeletal system, and hence a complete description is beyond the scope of this work. Preliminary introduction to human walking can be obtained through Inman [33] and, Rose [32]. Winter [34] provides a detailed analysis of kinematic, kinetic and muscle activation patterns of human gait. Nonetheless, a short description of relevant features and sub-modes of level walking is presented. While gait is essentially a 3-dimensional activity, most of the work is done in the sagittal plane [1]. In addition, since the prosthesis is constrained to operate in the sagittal plane, it also is the most relevant dimension. The angle conventions used to describe gait are shown in Fig. 3-1.

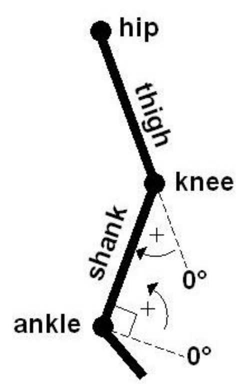


Figure 3-1. Joint Angle conventions for human leg.

A stick figure diagram of a complete gait stride (cycle) is shown in Fig 3-2, with accompanying joint angle behavior over a normal stride. The data is representative of normal walking gait of a healthy population [1]. Joint functionality within each gait stride can be adequately captured by dividing the cycle into four distinct modes as sectioned in Fig 3-3, each of which are described in further detail in subsequent paragraphs. Overall, a typical gait can be divided into stance (load bearing) and swing (non-load bearing) phase, each of which lasts approximately 60% and 40% of a gait cycle. A total of four functional sub-modes (two for stance and two for swing) are identified and described in subsequent sections.

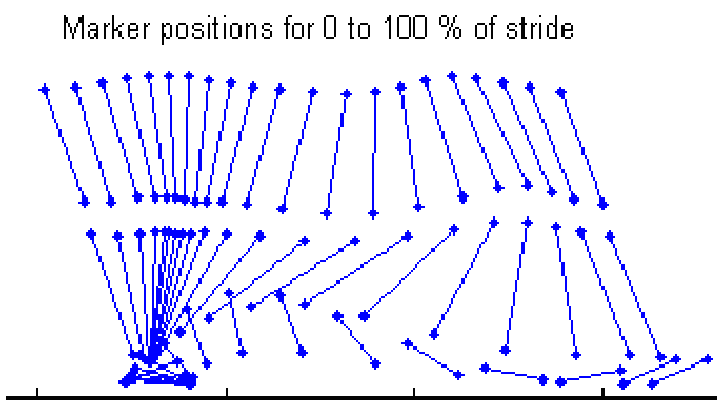


Figure 3-2. Stick figure sketch of human walking over a single stride cycle. A normal gait stride cycle is described as the time interval between consecutive heel strikes of the same leg.

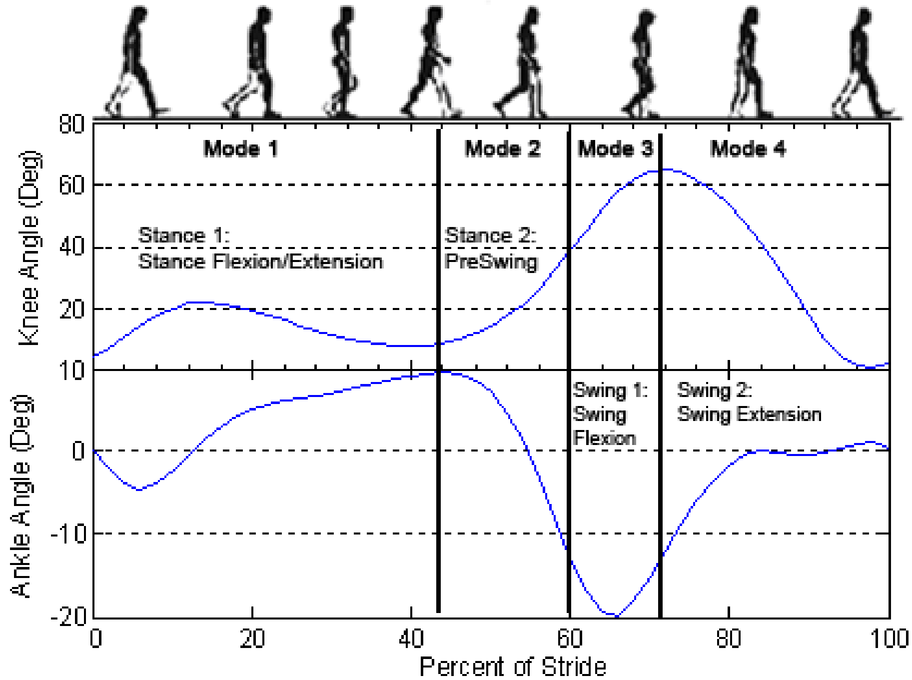


Figure 3-3. Knee and Ankle angle pattern over a typical gait cycle. Vertical lines represent the four distinct sub-modes within a gait period.

Four Gait Sub-Modes / States for a Normal Level Walk

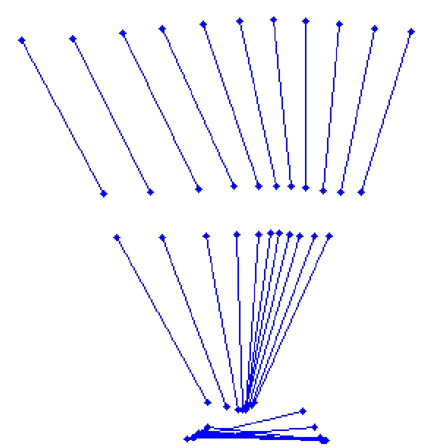


Figure 3-4. Stick figure sketch of Stance flexion / extension mode (0 – 40% of stride)

The Stance Flexion / Extension begins with heel strike upon which the knee immediately begins to flex so as to provide impact absorption and begin loading. The knee flexion during stance (stance flexion) increases considerably with speed and is the most noticeable kinematic indicator of speed. Immediately after heel strike, the ankle too begins to extend to reach flat foot. Once flat foot is reached, the ankle together with the knee provides extensive torque for weight bearing. As the knee extends towards the latter part of this mode, the ankle applies an increasing extensive torque in preparation for the push off phase of gait.

The Pushoff / Pre-Swing is the active power generation phase and begins as the ankle angle becomes extensive (i.e. ankle velocity becomes negative). The knee simultaneously begins to flex as the ankle plantarflexes to provide push off power. The push off combined with knee flexion also prepares the leg for the swing phase and hence is also referred to as “Pre-Swing”. The ankle and knee joint together work in a relatively much more coordinated manner during push off, so as to avoid any unnecessary jerks in the body, during power generation in the ankle.

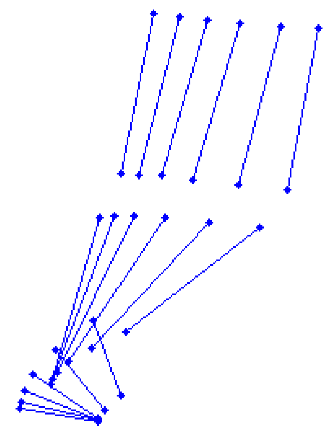


Figure 3-5. Stick figure sketch of Push off / Pre-Swing mode (40 – 60% of stride)

Early Swing (Swing Flexion) begins as the foot leaves the ground and lasts until it reaches a maximum flexion. The swing flexion is essential for proper ground clearance and increases with speed. However, the change in swing flexion with speed is much less to that compared to stance flexion. Individuals with gait disorders such as stiff-knee gait often exhibit reduced swing flexion [35], making them susceptible to stumbling and other problems. Healthy maximum swing flexion angles usually lie between 60-70 degrees.

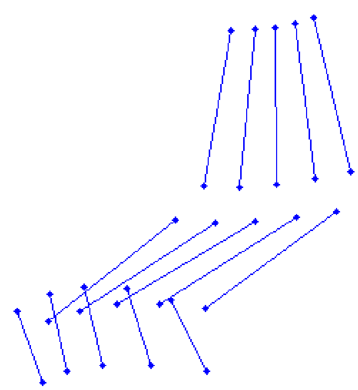


Figure 3-6. Stick figure sketch of swing flexion mode (60-73%) of stride

Late Swing (Swing Extension) mode represents the forward swing of the leg, when the limb swings forward in anticipation of heel strike. The muscles are relatively dormant during early part of the swing, with larger braking torques in the knee joint towards the end to avoid impact during full extension. After reaching full extension, the knee flexes slightly in anticipation of heel strike.

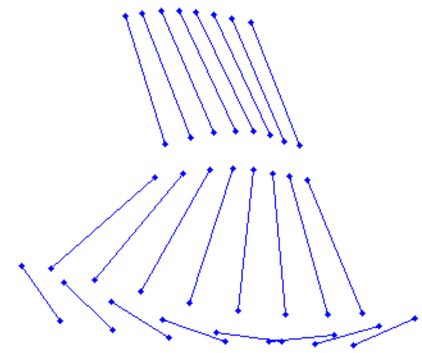


Figure 3-7. Stick figure sketch of swing extension mode (73-98%) of stride

The ankle torque remains relatively dormant during the swing phase, as it is not involved in the forward swing of the leg and remains fairly neutral (close to zero degrees) in anticipation of heel strike The knee joint in swing phase during slow and normal walking is largely passive phase (Fig. 1-1) [1], since it only aids with stopping the knee, as the natural dynamics of the body are usually sufficient in aiding the shank swing.

CHAPTER IV

THE CONTROLLER

Unlike any prior work, this paper describes a method of control for a powered knee and ankle joint together that enables natural, stable interaction between the user and the powered prosthesis. The controller is similar to prior works described above in that it divides gait into sub-modes or finite states and uses biomimetic characteristics to the maximum extent possible. However, the overarching approach in all prior work (on the control of active joints) has been to generate a desired joint position trajectory, which by its nature utilizes the prosthesis as a position source. Such an approach poses several problems for the control of powered transfemoral prosthesis. First, the desired position trajectories are typically computed based on measurement of the sound side leg trajectory, which 1) restricts the approach to only unilateral amputees, 2) presents the problem of instrumenting the sound side leg, and 3) generally produces an even number of steps, which can present a problem when the user desires an odd number of steps. A subtler yet significant issue with position-based control is that suitable motion tracking requires high output impedance, which forces the amputee to react to the limb rather than interact with it. Specifically, in order for the prosthesis to dictate the joint trajectory, it must assume a high output impedance (i.e., must be stiff), thus precluding any dynamic interaction with the user and the environment.

Unlike prior works, the approach proposed herein utilizes an impedance-based approach to generate joint torques. Such an approach enables the user to interact with the prosthesis by leveraging its dynamics in a manner similar to normal gait [31], and also generates stable and predictable behavior. The essence of the approach is to characterize the knee and ankle behavior with a series of finite states consisting of passive spring and damper behaviors, wherein energy is delivered to the user by switching between appropriate spring stiffness (of the virtual springs) during state transitions or by actively manipulating the equilibrium point (of the virtual springs). In this manner, the prosthesis is guaranteed to be passive within each gait mode, and thus generates power simply by switching between modes. Since the user initiates mode switching, the result is a predictable controller that, barring input from the user, will always default to passive behavior.

Though very apt and versatile, the human locomotion system is not a precise mechanism like a robotic manipulator. The variability in joint torques from cycle to cycle during gait indicates an adapting human locomotive system. In fact, Winter [31] reports that while the sum of hip, knee and ankle joints over a gait stride have consistent repeatability, the individual torques in the hip and knee joint have great variability. This demonstrates that any external prosthesis needs to provide sufficient compliance for a more natural interaction while delivering required torques. In addition, it is hypothesized that the central nervous system often attempts to use the natural dynamics of the body to accomplish physical tasks. By concretely defining the dynamic behavior of the prosthetic joints, a better integration between the prosthetic user and the device can be expected. If optimal joint dynamic behavior can be captured and consistently delivered to the user, it becomes a natural extension of the body instead of an additional device such as wheel chair or crutch that only has certain functional value. An impedance-based approach allows for this dynamic characterization of knee and ankle joint, which in addition to providing the required joint torques also provides a predictable behavior under different joint configurations.

CHAPTER V

IMPEDANCE CHARACTERIZATION OF GAIT

Based loosely on the notion of impedance control proposed by Hogan [36] the torque required at each joint during a single stride (i.e., a single period of gait) can be piecewise represented by a series of passive impedance functions. A simple linear spring damper model is proposed to characterize joint behavior within each gait mode, and a preliminary regression analysis (see Appendix A) of gait data for normal walking (see Fig 5-1) from Winter [1] indicates that joint torques can be approximated to a good degree as functions of joint angle and velocity by the simple impedance model

$$\tau = k_1(\theta - \theta_e) + b\dot{\theta} \tag{1}$$

The linear stiffness and damping are described by k1 and *b*, θ_e is the equilibrium angle or the set point, and the angle, θ and torque, τ, are defined as in Fig. 3-1. If the coefficients *b* and k_1 are constrained to be positive, then the joint will attempt to converge to a stable equilibrium at $\theta = \theta_e$ and $\dot{\theta} = 0$ within each gait mode. In addition, during certain modes, where the joints act purely in a braking manner, non-linear elements can be introduced to produce appropriate torque profile as long as the torque is constrained to be opposite in sign of the joint velocity. Thus, in any given mode, the behavior is passive, and will come to rest at a local equilibrium, thus providing a reliable and predictable behavior for the human user.

The characterization of the control framework by relating torque to angle and velocity states means the angular profile of the prosthesis need not be explicitly described in time.

This avoids the necessity of trajectory generation, which can be a cumbersome task to define for all possible subsets of locomotion. In addition, specifying the position profile of the prosthesis means the user needs to adapt to the prosthesis instead of the other way around, which likely could increase rehabilitation time.

Effective impedance based characterization becomes possible due to segmentation of gait into finite modes (see Fig 5-1) exhibiting distinct behaviors, as described in the earlier sections. Such segmentation is made possible by analyzing torque phase space data (torque-angle relationship), and identifying appropriate transitions between modes (see Finite state Fig 8-1). Since the user initiates mode changes, the prosthesis barring input from the user is constrained to a certain mode and thus to a certain dynamic behavior. This is likely to facilitate better control and predictability of the device to the user.

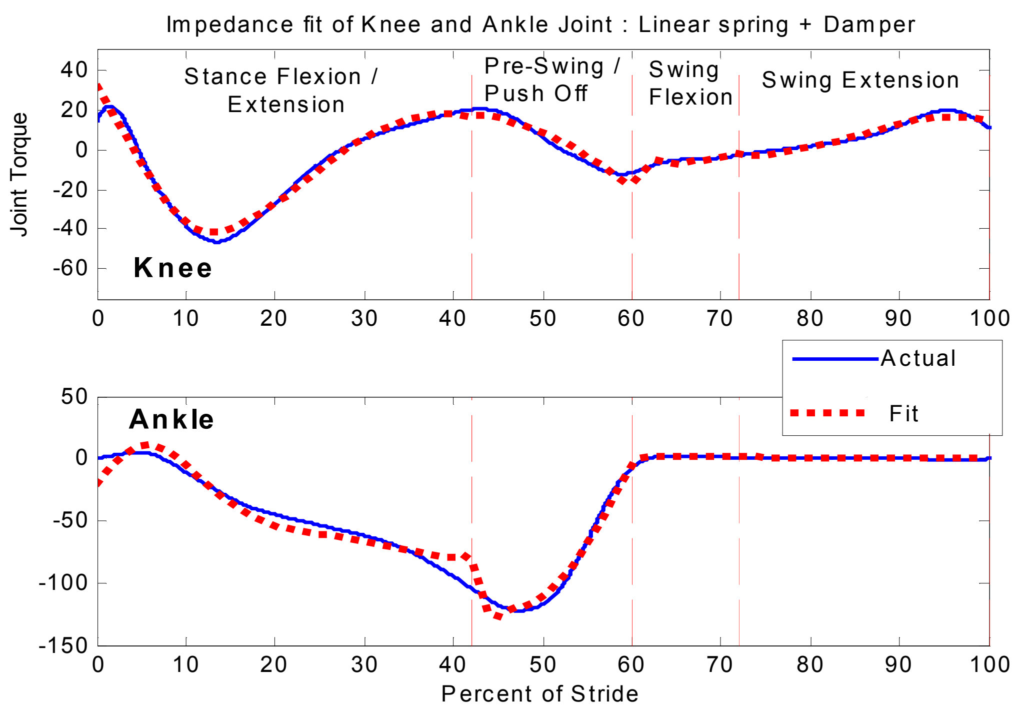


Figure 5-1. Piecewise fitting of knee and ankle torques during normal speed level walk (averaged population data from Winter, 1991 scaled for a 75 kg adult) to a spring-damper impedance model. The vertical lines represent the segmentation of a gait stride into four distinct modes.

While a simple spring damper model provides an effective starting point to generating necessary restoring torque, certain phases such as knee break during pre-swing, and the stiffening of the ankle torque prior to push off, require more than the simple model
In such cases, the impedance model is augmented by altering the set-point angle of the virtual spring in a dynamic manner. Such behaviors are described in a detailed analysis of mode segmentation later in the thesis. In addition, though not shown in Fig 5-1, a supplementary mode is added during early stance in the ankle joint to enable smooth transition during heel strike.

CHAPTER VI

MOMENT OF SUPPORT CHARACTERIZATION

Transfemoral amputees often compensate the loss of knee and ankle joint by an extra input in the hip joint [1]. In normal adults, the walking load is shared between various joints. However, the sharing of joint loads during stance varies between individuals and even between strides. As reported by Winter [31], while the algebraic sum of the hip, knee and ankle joint (referred to as 'Moment of Support') is consistently positive and repeatable, individual knee and hip joint even in normal adults show highly variable patterns during stance period of normal walking (the ankle joint relatively is much more consistent). While this observation is of qualitative nature, it does have significant relevance to the working of transfemoral prosthesis. For instance, what specifically should the torque at the knee joint be is a function of hip activity as well. Since, the user is likely to undergo sufficient rehabilitation, it is assumed that this problem will be solved in reverse i.e. the user will learn to recognize what hip moments are needed to interact in a useful manner with the prosthesis as long as the prosthesis provides sufficient support. This lends further support to the necessity of consistent behavior on part of the prosthesis. In addition, it appears that as long as the knee torque is within a sufficient regime of values, gait is facilitated (though ankle torque needs to be much more accurate). Hence, precise computation of torque values may not be needed; rather simply approximation of joint dynamics is likely to suffice and is the approach taken here. In the following chapters, we infer on a suitable set of joint dynamic behavior based on biomechanical data [1] of a healthy population with the intent to find general relationships that will allow for the generation of sufficient joint torques to facilitate basic gait in a stable manner. Since, the idea is to establish a basic framework; no attention is paid to optimizing gait variables such as energy efficiency, workload in hip joints etc.

CHAPTER VII

IMPEDANCE CHARACTERIZATION OF BIOMECHANICAL GAIT

The knee and ankle have very different roles during the stance and swing phase of gait. Since the ankle joint is directly in contact with the ground, it is the primary joint of interaction with the external environment and hence dictates the overall energy input into the body during walking. The ankle provides positive power while simultaneously also guarding against collapse at the ankle joint. It is thus constrained to produce an extensive torque through most of stance. The knee on the other hand, has a dual task of supporting the body by providing sufficient torques at the joint, while also helping to ensure that vertical trajectory of the body undergoes minimum diversion. During swing phase the knee is primarily involved in limb advancement by swinging the shank through, as the ankle joint remains relatively dormant. Though separate in their functionalities they do interact significantly to facilitate forward movement in a smooth, stable manner.

In subsequent paragraphs, impedance behavior of the knee and ankle joint is approximated from mass normalized population data [1]. Four functional modes or finite states (Fig 5-1) of level walking are described in detail for both the knee and ankle joint. As mentioned in the previous discussion, the goal here is to identify general relationships, that “sufficiently reproduce” joint behavior. Attempt is made to validate the model against three different speeds: slow, normal and fast, though the data for ‘normal’ is more dominantly referred to.

Stance Flexion / Extension, 0% - 40%

<u>Knee Joint</u>

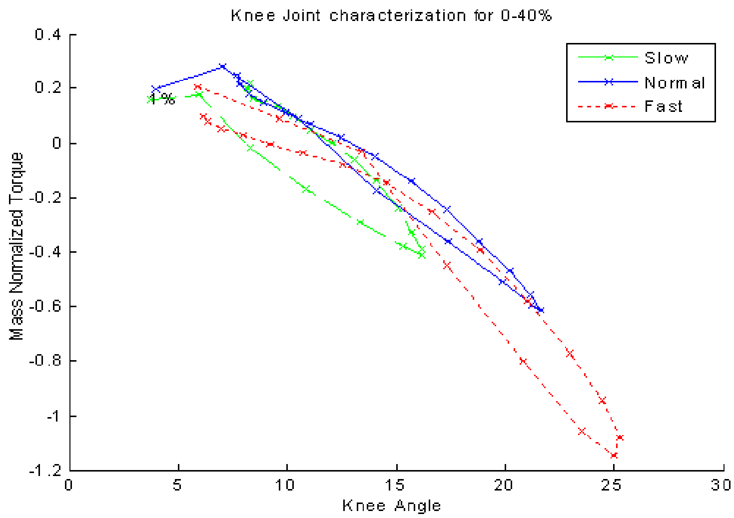


Figure 7-1. Mass normalized Torque-Angle phase space relationship in the knee joint during Stance flexion / extension mode (0 – 40%) of stride

During initial stance period the knee goes through a flexion-extension phase that serves to absorb impact after heel strike, and to prevent unwanted vertical movement of the trunk. As can be seen from the torque-angle phase space (Fig 7-1), there is dominant linear relationship between the knee torque and angle. The characterization is further improved by a damper that accounts for the hysteresis type loop behavior. The parameters for the mode are determined by a least squares fit to be:

Table 7-1: Impedance model parameters for stance flexion / extension phase of knee joint shown in Fig 7-1

	K1	b	Set-Point
Fast	0.0635	0.0003	10.0000
Normal	0.0557	0.0003	12.0000
Slow	0.0604	0.0019	11.0000

The stiffness coefficients remain fairly constant through the three walking speeds and are likely to explain the significant increase in stance flexion with speed. As the body moves faster during walk the trunk has a greater momentum during heel strike thus causing the knee joint to flex more. If the knee joint responded to higher speeds by a stiffer knee, then it would oppose the natural dynamics that tended to cause knee flexion to a greater degree.

Reconstructed torques from the estimated parameters indicate a relatively better fit for slow and normal walking, though the general relationship is still visible in fast walking as well. If deemed necessary during actual experiments, the fitting could be further improved upon by adding an extra cubic term that maintains passive behavior while providing a better fit, but simplicity was preferred given the necessity for experimental tuning. The positive set point indicates that the knee joint initiates knee flexion by exerting a flexive torque during early stance, and also helps the transition into push-off, by generating a flexive torque towards the end of stance. Though, acting independently such flexive torques would cause the knee to buckle; during gait, they help to counteract the extensive torque exerted by the hip on the knee joint. This provides vertical support while avoiding hyperextension in the knee.

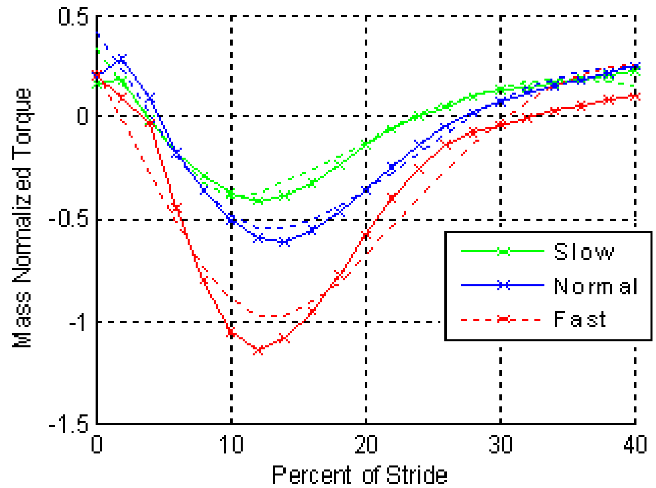


Figure 7-2. Comparison between actual torques in the knee joint (Solid lines) with torque reconstructed from estimated parameters (Dashed lines) during stance flexion / extension mode (0 – 40%) of stride

<u>Ankle Joint</u>

The ankle joint exhibits two different functionalities during the stance flexion/extension period. Immediately following heel strike, the angle applies a dorsiflexive torque, to ease the foot into the ground and avoid any slapping (as see in drop foot gait). This behavior last for a very short period (0% - 6%), until the foot touches the ground and ankle angle begins to flex. The ankle stiffness is relatively less compared to later in the stance to allow for more compliant interaction with the ground.

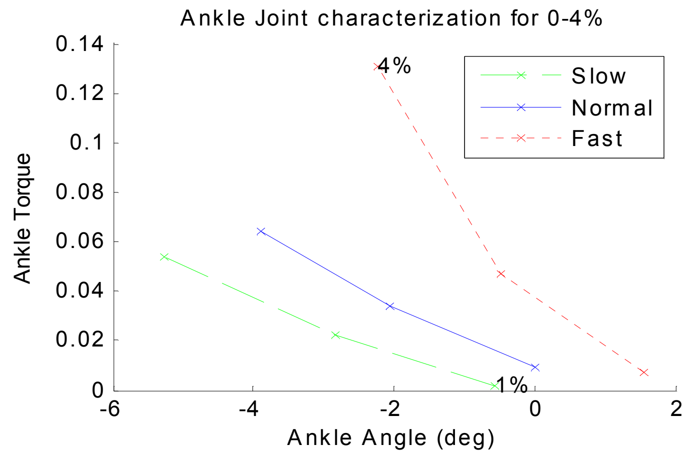


Figure 7-3. Mass normalized Torque-Angle phase space relationship in the ankle joint during early part of Stance flexion / extension mode (0 – 4%) of stride

Table 7-2: Impedance model parameters for early part of stance flexion / extension phase of ankle joint shown in Fig 7-3

	K1	b	Set-Point (deg)
Fast	0.0325	0.0002	1
Normal	0.0125	0.0003	1
Slow	0.0105	0.0001	-1

Once the foot reaches the ground, the loading of the ankle begins. The torque-angle space behavior (Fig 7-4) initially exhibits a linear relationship between torque and angle, and this plantarflexive torque is likely aimed at preventing buckling at the ankle joint. Once, the ankle joint is flexed past a certain threshold, the ankle produces an increasingly larger force that increases non-linearly until the joint produces sufficient torque to begin extension. This onset of non-linear behavior could in some sense be regard as the beginning of push off, since the amount of torque the persons responds with at this phase plays a significant role in determining the speed of the person. One notable feature of the ankle torque during this phase is that for most of period, the torque is higher for slower speeds, which could be considered counter-intuitive during first glance. However, less flexive resistance in the ankle joint allows the shank to rotate ahead fast and more easily.

Later during the unloading or propulsive push off we will observe higher flexive torques with faster speeds.

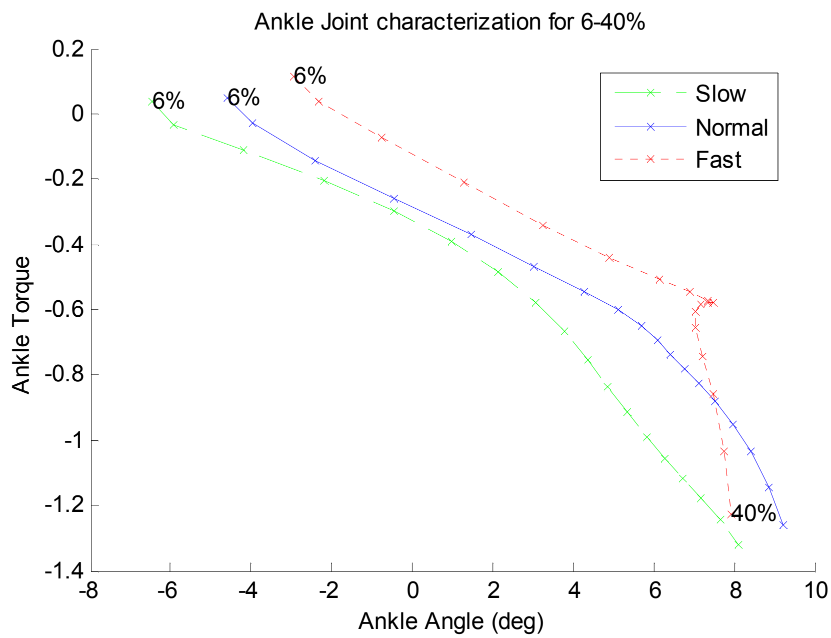


Figure 7-4. Mass normalized Torque-angle relationship in the ankle joint for stance flexion-extension (6-40%) phase

The overall ankle behavior is captured by initially determining the stiffness of the ankle during the linear phase (see Fig 7-5). Once the threshold is crossed, the non-linear increase in the force is characterized as a non-linear movement of the set point, while the stiffness remains constant. The stiffness (Table 7-3) in the ankle appears slightly larger with speed, though the set point of the ankle spring ensures that smaller torques are produced so as to minimally impede joint motion without allowing buckling.

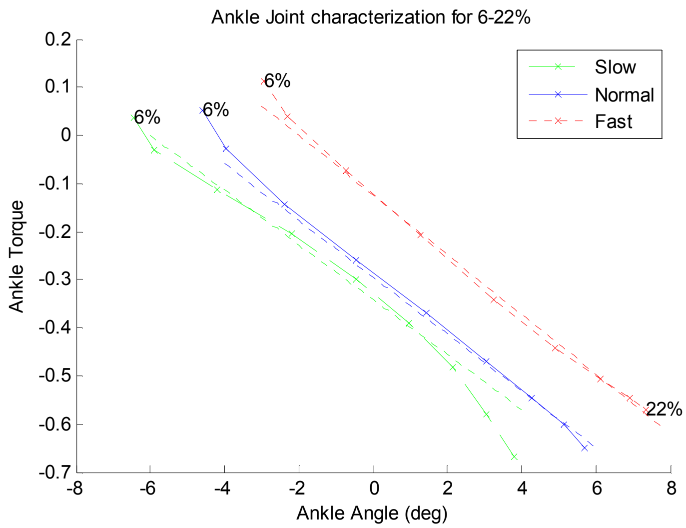


Figure 7-5. Linear mass normalized torque-angle relationship in the ankle during middle of stance flexion-extension mode (6-22% of stride). Solid lines indicate actual data, while the fit is represented by dashed lines.

Table 7-3 Stiffness parameters for linear part of stance flexion / extension phase of ankle joint shown in Fig 7-5

	K1	**Set-Point (deg)**
Fast	0.0621	-2
Normal	0.0590	-5
Slow	0.0569	-6

The movement of the ankle set point (Fig 7-6.) shows contrasting behavior at different speeds. At slow speeds the change in set point is initiated at less ankle flexion, and increases at a slower rate. In contrast, the set point change during fast walking happens with no change in ankle angle. This sudden movement of the set point could be explained through sudden release of the contracting muscle to generate a propulsive torque. The overall principle of torque generation in the ankle during this mode is characterized by the following principle using a power fit to describe the movement of the set point:

$\tau = k_1(\theta - \theta_e)$, Where, $\theta_e = \theta_0$ for $\theta \le \theta_{threshold}$

$$\theta_e = A \cdot (\theta - \theta_{threshold})^{\lambda} + \theta_0 \quad \text{for} \quad \theta > \theta_{threshold}$$

Where, θ_0 And k_1 are the set point and stiffness value characterized in Table 7-3. A fit for the movement of the set point is shown in Fig 7-5. The fit parameters are stated in Table 7-4.

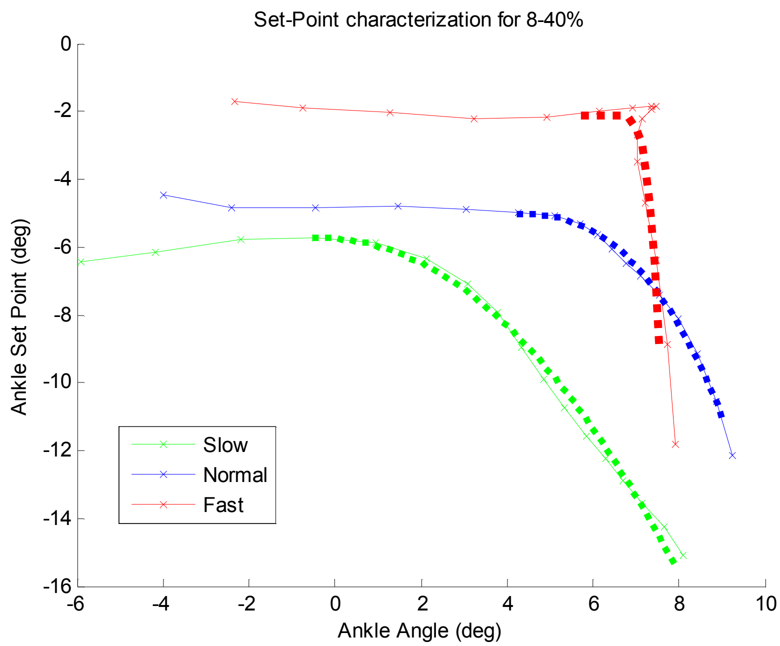


Figure 7-6. Characterization of movement of set point of the virtual spring at the ankle joint during for stance flexion-extension (6-40%) phase. Solid lines indicate actual data, while dashed lines represent the fit.

Table 7-4. Power fit parameters for the non-linear movement of set point of the virtual spring during stance flexion-extension (6-40%) phase as shown in Fig 7-6.

	A	λ	**Threshold**
Fast	-0.1145	7.729	-6
Normal	-0.1145	2.517	-4
Slow	-0.1145	2.089	0

The choice of a power fit to represent the movement of the set point has a few attractive features. The speed can be characterized by varying 'λ' and threshold, both of which are very intuitive, since one almost needs to be adjusted in inverse of the other. The coefficient ‘A’ yields a good fit at a given constant value and need not be adjusted if the tuning is limited to the aforementioned two variables. In addition, distinctly different curves representing very different behaviors can be generated only by slight adaptation of the tuning variables.

The power fit characterization of the set point does not lend itself to the normal passivity characterization of the spring damper model. However, the passive behavior of the ankle is still guaranteed since the set point movement is limited to the mode where the ankle torque is negative while the angle is flexing (positive velocity), thus constraining the ankle power to be negative. As soon as the movement of the set point causes the ankle velocity to become positive (extensive) and the ankle power becomes positive, the ankle joint switches onto the next mode, where the spring-damper joint characterization is resumed. Thus, the introduction of the power curve only allows for a better characterization of non-linear ankle torque characterization without risk of instability.

Pre-Swing / Pushoff, 42% - 60%

Knee Joint

Once the ankle becomes extensive the knee simultaneously begins to flex to avoid unnecessary excursions in the trunk while the ankle generates forward propulsive power through active extension in the joint. The coordination between the ankle and knee joint is relatively higher during this mode. In fact, the knee and ankle angle vary linearly with respect to each other during this mode, irrespective of the walking speed (See Fig 7-7). The ratio of the joint angles is approximately 1.4 during this phase, and remains relatively unchanged during different tasks. This is indicative of the high degree of positional coordination necessary for the proper execution of the push off phase.

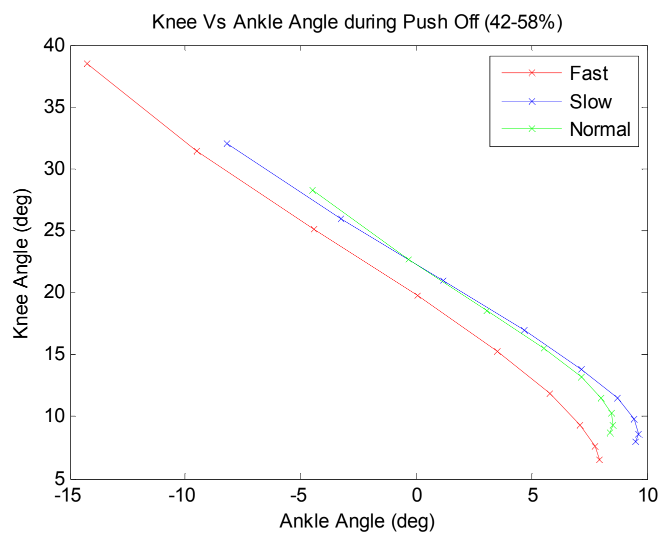


Figure 7-7. Ratio of Knee and Ankle angle during push off

The knee torque-angle behavior for this mode is shown in Fig 7-8. The knee joint is in maximum extension towards the end of the stance flexion-extension mode, and begins knee break by initiating flexion through a flexive torque. Once, the knee flexes beyond the set point and the knee torque becomes extensive (negative), the joint behaves as a non-linearly softening spring (the ankle in contrast behaved as a hardening spring during the earlier stance phase).

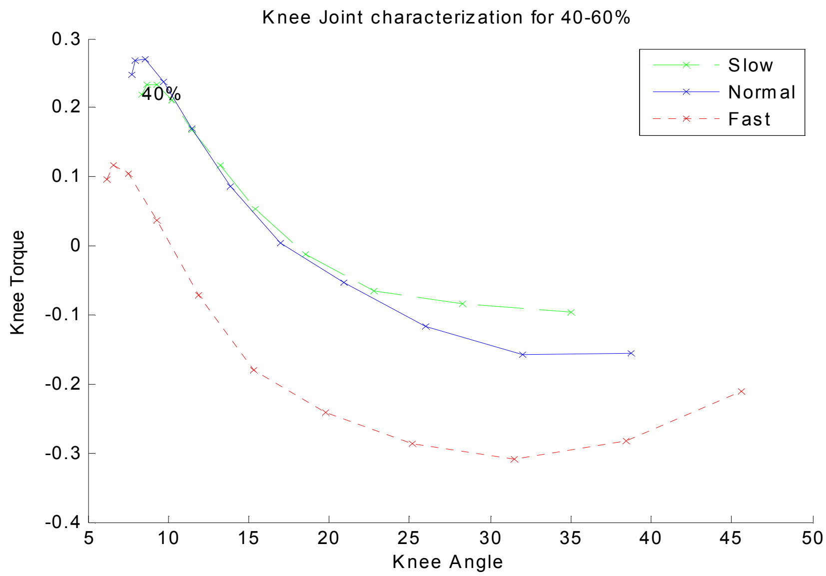


Figure 7-8. Mass normalized torque-angle relationship in the knee joint during Push off mode (40-60% of stride).

The characterization of knee torque during phase space is done by modeling a spring with a set point dependent on the knee angle. The knee stiffness is specified during the initial flexive torque behavior and is held constant through the entire time. The stiffness is for all practical purposes uniform over various walking speeds, and it is only a differing set-point that distinguishes the torque behavior during various walking speeds

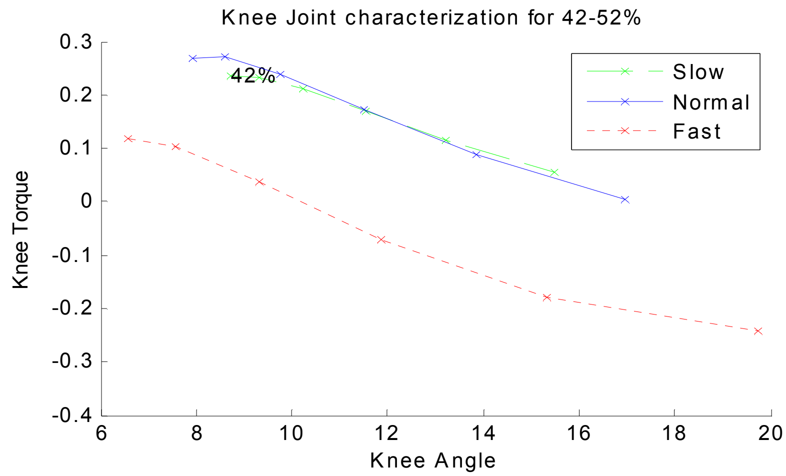


Figure 7-9. Linear mass normalized torque-angle relationship in the Knee during Push off (42-52% of stride).

Table 7-5. Stiffness parameters for linear part of Push off phase of knee joint shown in Fig 7-9

	K1	**Set-Point (deg)**
Fast	0.037	10
Normal	0.0311	17
Slow	0.0300	17

The movement of the knee set point under the assumption that the stiffness is constant for various speeds is shown in Fig 7-10. The movement of the set point is modeled as a quadratic function of the knee angle after the knee torque becomes extensive. It is important to note that passivity of the controller would not be maintained under a quadratic relationship if the knee torque were positive since the joint would produce positive power (note that the knee velocity is positive). However, since the set point is moved in a quadratic relationship after the knee torque becomes extensive (i.e. negative), the joint is limited to negative power at the joint, and thus constrained to be passive. The final knee joint behavior for this mode can be characterized as:

$\tau = k_1(\theta - \theta_e)$, Where, $\theta_e = \theta_0$ for $\theta \le \theta_0$

$$\theta_e = 0.0153(\theta - \theta_0)^2 + 0.424(\theta - \theta_0) + \theta_0 \text{ for } \theta > \theta_0$$

The first two coefficients of the quadratic function of the set point describe the shape of the curve, which is then appropriately scaled by the original set point. Essentially, once a suitable stiffness value is experimentally determined, simply choosing a suitable set point allows for variation in walking speed.

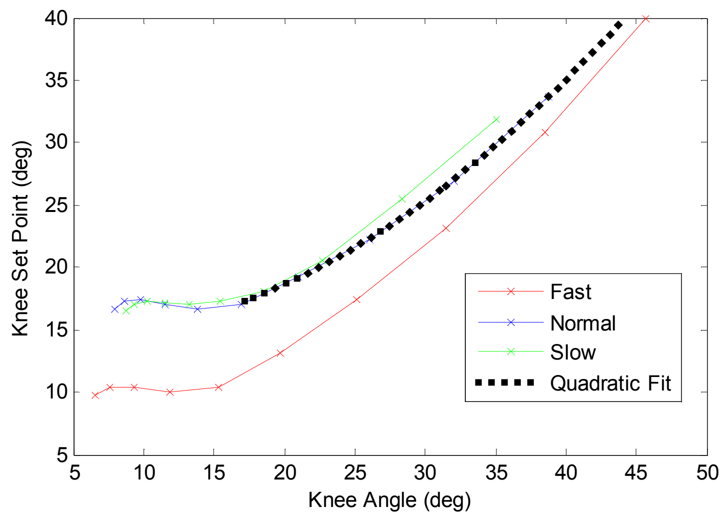


Figure 7-10. Characterization of movement of set point of the virtual spring at the knee joint during Push off phase (40-60%) phase. Solid lines indicate actual data, while dashed lines represent the fit.

Ankle Joint

The push off phase in the ankle joint begins when the ankle becomes extensive. This is the unloading phase in the ankle joint, as the ankle provides positive power to aid push off. The torque space behavior of the ankle joint during this phase is fairly linear and hence lends itself well to characterization as a linear spring (see Fig 7-11). The parameters (Table 7-6) indicate that the set point rather than the stiffness needs to be adjusted to enable different speeds.

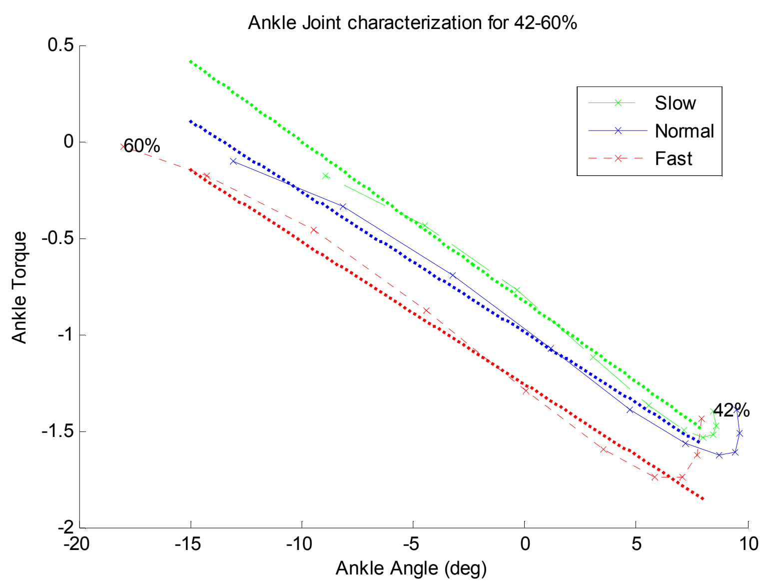


Figure 7-11. Mass normalized torque-angle relationship in the ankle joint during Push off mode (42-60% of stride). Solid lines indicate actual data, while dashed lines represent the fit.

Table 7-6. Linear stiffness of the ankle joint during Push off mode shown in Fig.7-11

	K1	Set-Point (deg)
Fast	.083	-17
Normal	.073	-13.5
Slow	.074	-10

The discussion of joint behavior during swing phase is limited to the knee joint since it has a more dominant role during this phase. The ankle torque during swing is small and is simply characterized as a weak spring with a set point close to zero degrees. The knee on the other hand swings back to reach a maximum swing flexion angle before swinging forward to near full extension in anticipation of heel strike. Previous controller for variables impedance knees [9, 28] have characterized swing phase as an entirely passive mode. Human gait data does indicate the swing phase as primarily passive, where the natural dynamics are sufficient to induce swinging, and the joint primarily works as a brake.

The inertial properties of the human leg and the prosthesis are different, and this becomes very apparent during swing. During stance the overall forces imposed on the prosthesis are much higher, and the prosthesis is in a higher output impedance mode, thus precluding any influence of the physical dynamics of the prosthesis. However, during swing, a human leg is able to swing fairly well and the impedance at the joint is small. Comparatively, the back drivability of pneumatic actuators impedes free swinging motion of the prosthesis, though the issue is partly corrected by adding feed forward velocity compensation, human like low impedance is still difficult to achieve. Hence, the analysis of knee behavior during swing using biomechanical data may not be directly applicable to a prosthesis. Nonetheless, a simple spring damper model is still found to be fairly close to actual biomechanical data and discussed in subsequent as a feasible model.

Swing Flexion (60 – 72 %)

Swing flexion begins immediately after toe off, as the shank swings back while the hip is swinging forward. A steady extensive braking torque causes the knee joint to achieve some amount of maximum flexion before the forward swing. This flexion angle often referred to as “maximum swing flexion” angle is critical to achieving ground clearance, since it allows the hip to travel sufficiently forward before the shank swings through. A healthy swing flexion for normal adults is usually between 60-70 deg and increases with walking speed [1]. The simplest characterization of knee joint in this mode is a simple damper. This has the advantage of presenting only a single tuning variable (the damping coefficient) and a single measure of the outcome (maximum swing flexion angle), with a clear defined relationship between the two.

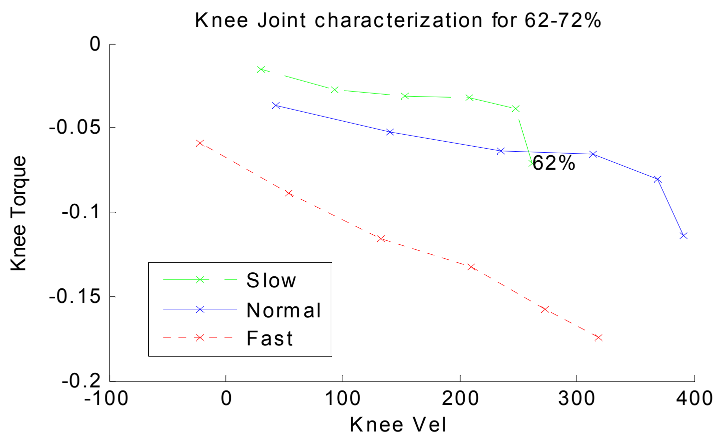


Figure 7-12. Mass normalized torque-angle relationship in the knee joint during swing flexion mode (62-72% of stride). Solid lines indicate actual data, while dashed lines represent the fit.

Table 7-7. Linear damping parameters of the knee joint during Push off mode shown in Fig. 7-12

	Damping Coefficient
Fast	644e-6
Normal	238e-6
Slow	174e-6

The biomechanical data does display a linear damping (Fig 7-12), and the damping values are higher for fast walking speeds (Table 7-7). This increase in damping protects the knee joint from flexing to larger than desired angle as the joint velocity increases with increase in walking speed. If desired flexion is not achieved even when the damping is set to zero, then an additional spring element can be added with a set point less than the desired swing flexion so that it provides some amount of positive power to swing the leg up to the desired angle. However, in most prosthesis with variable impedance mechanisms, velocity dependent damping are the norm and have appeared to function well [9, 28], though they can be powerless if under minimum friction the desired swing flexion is not achieved [28]

Swing Extension (72-98%)

Once the maximum swing flexion is achieved, the leg swings forward until it reaches close to full extension. The braking in the knee joint during forward swing is fairly non linear (Fig 7-13). The torque remains low initially allowing for a free swing and increases rapidly as the leg approaches full extension. Due to the stiff back drivability of pneumatic actuators and limited bandwidth, a free swing of the prosthesis is hard to attain (even with feed forward velocity compensation) and hence the normal biomechanical profile (Fig 7-13.) may be not be applicable. However, the goal in the forward swing is fairly well defined, as the leg has to swing from a given initial angle to close to full extension in some given time period in a smooth manner. As an alternative solution, a simple spring and damper model is used, and pure damping behavior during swing is replaced by an initial injection of power in the beginning of the swing that is later braked by the spring and damper together to avoid impact at full extension. The stiffness of the joint is tuned to adjust the time the leg takes to swing through (as stiffness is directly related to frequency in a mass-spring-damper system) and the position of the spring set point is adjusted to stop the knee angle at the desired value. Hence, the pure passive functionality of the knee joint in normal gait, is replaced by a simple spring damper that provides easy tunability.

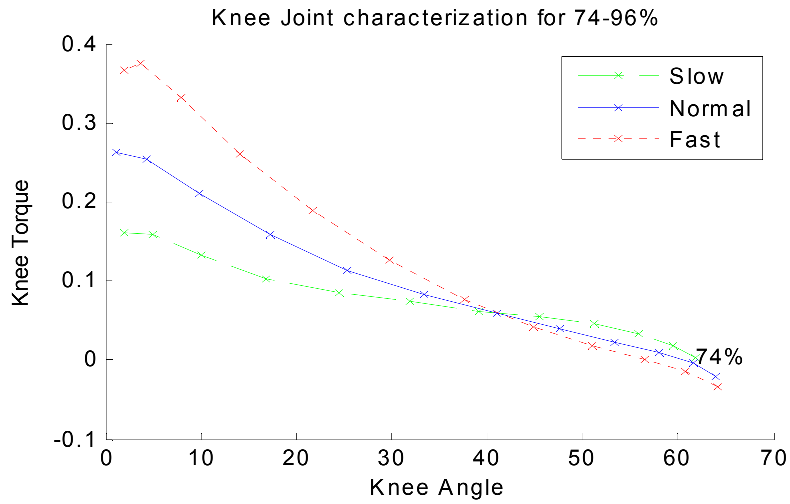


Figure 7-13. Mass normalized torque-angle relationship in the knee joint during swing extension mode (74 – 96 % of stride).

CHAPTER VIII

CONTROLLER IMPLEMENTATION AND TESTING SETUP

The finite state impedance-based controller was implemented using Matlab (7.3) and Simulink (6.5). The gait algorithm was implemented as a Finite state machine in Simulink using State flow model (Fig 8-1). The various gait modes described are each treated as an individual state and appropriate transitions between the two are defined. The overall Simulink implementation is available in Appendix B. As this is meant to be a preliminary implementation of level walking at a pre-defined speed, extra transitions between gaits are not included to avoid unnecessary chattering as can happen with a switching control algorithm. A proportional control method was implemented to provide low level force control in the actuator and gains were experimentally tuned to allow best force tracking in a stable manner. A National Instrument Data Acquisition card (PCI-6031E) was used to electrically interface to the tethered prosthesis.

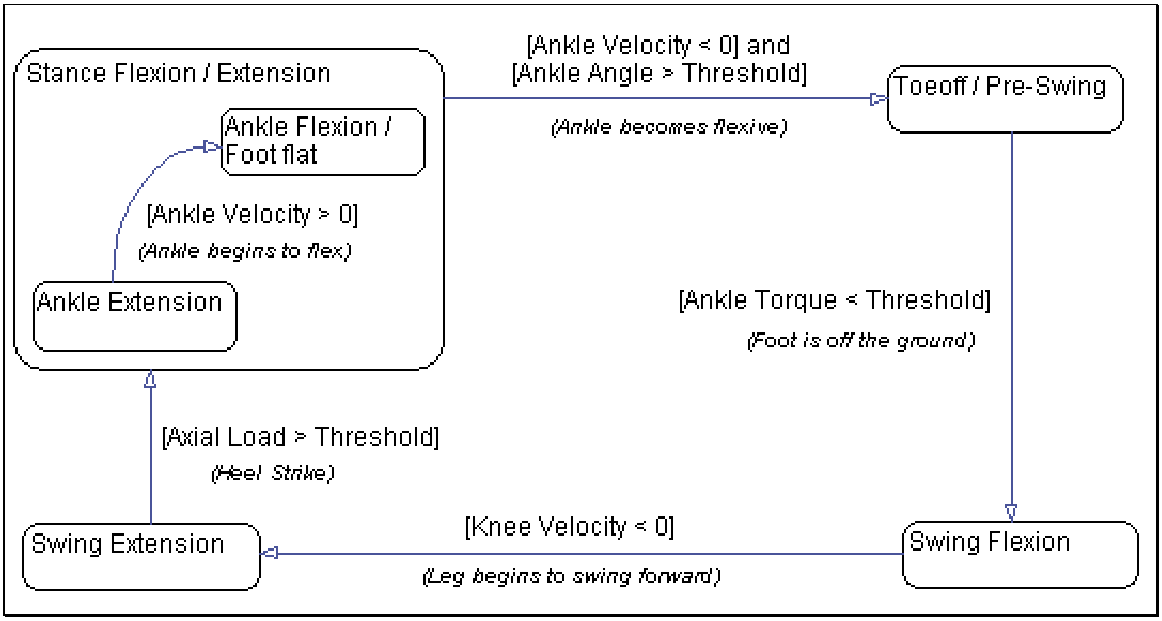


Figure 8-1. General Finite State model of gait. Each box represents a possible state and the appropriate transition between states are specified

The gait control strategy was implemented on the tethered prosthesis prototype on a healthy subject using an able-bodied testing adaptor as shown in Fig. 8-2. The adaptor consists of a commercial adjustable locking knee immobilizer (KneeRANGER-Universal Hinged Knee Brace) with an adaptor bracket that transfers load from the subject to the prosthesis. Since the prosthesis remains lateral to the immobilized leg of the healthy subject, the adaptor simulates transfemoral amputee gait without geometric interference from the immobilized leg. While the adapter allows for preliminary testing of the gait control algorithm, the setup does involve certain drawbacks in simulating prosthetic gait, some of which include 1) compliance of the soft tissue interface between the device and user (more so than exhibited by a limb/socket interface), 2) “parasitic” inertia of the intact lower limb (i.e., in addition to the inertia of the prosthesis), and 3) asymmetry in the frontal and axial planes which results in a larger than normal planar moments (i.e., as seen in Fig 8-2). Despite these, the adaptor provides a reasonable facsimile of amputee gait, and enables testing of the device and proposed impedance-based control approach.

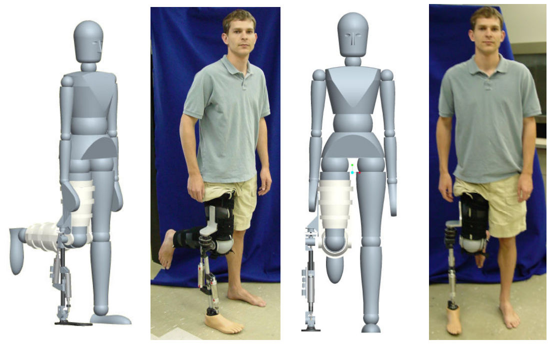


Figure 8-2. Able-bodied testing adaptor for enabling development, testing, and evaluation of the prosthesis and controllers prior transfemoral amputee participation.

Gait trials were performed on a treadmill, which provided a controlled walking speed and enabled enhanced safety monitoring, including a safety suspension harness and the use of handrails. The prosthesis was provided a 2Mpa (300 psig) pressure supply through a tether. Biomechanical impedance parameters estimated from population data [1] were used as the starting point and iteratively tuned through feedback from the user, recorded gait profiles and video to produce a gait pattern that seemed satisfactory to the user and showed reasonably good gait profile.

CHAPTER IX

EVALUATION AND TUNING OF PROSTHESIS

The quality of gait can be assessed in both qualitative and quantitative terms. Foremost, the comfort and feedback of the user needs to be taken into consideration and is also the most useful in evaluating the contribution of a prosthetic device. The user is often perceptive to power input from the device (e.g. ankle push off torque) and its overall dynamics. In addition, a simple visual assessment of gait can reveal obvious gait pathology such as excessive and asymmetrical movements in the body. While qualitative analysis provides a very reasonable and appropriate method of evaluating gait, it does have certain limitations. For instance, gait trials lasting a prolonged period allow the user to adjust to different prosthetic behavior, and this can affect the user feedback. A user may easily get accustomed to working with less than desired push off at the ankle joint or non-ideal behavior during heel strike, and thus may not provide adverse feedback once he has adapted to these disturbances in the device. In addition, visual assessment, even by experts, cannot be relied on consistently and even when a problem is visually observed the cause is difficult to determine [37]. Hence, it is necessary to have some amount of quantitative variables that can be consistently measured to assess the quality of gait without reliance on human intuition.

While a single quantitative measure of the quality of gait is hard to define, certain pointers do exist as guides. Quantitative analysis of gait can be done by analyzing temporal-spatial, kinematic, kinetic and energy related parameters [38]. The overall data that can be collected for evaluation purposes are limited by the set of sensors available in the laboratory setting. The sensor set used are as follows:

(a) Knee and Ankle Potentiometer

(b) Load sensors to measure joint torques

(c) Socket load cell above prosthetic knee together with footswitches to detect heel strike and toe off events on prosthetic and sound side.

A clinical assessment of gait of an amputee is likely to involve a more comprehensive measurement of gait parameters. However, since this experiment constitutes a preliminary study of assessment and tuning of prosthesis, only a minimal set of parameters are chosen so as to keep the entire process compact and manageable. The goal of the entire tuning and gait evaluation process is to (a) improve temporal symmetry; (b) reproduce healthy looking kinematic data and (c) Minimize jerks and significant physical asymmetry in walking. Each of these domains is discussed in further detail in the following sections.

TEMPORAL MEASURE

While amputees markedly walk at slower self-selected speeds, the treadmill constrains the user to a pre-selected speed. However, the stance and swing times on normal and prosthetic leg spent to achieve the desired speed may not be symmetrical. In fact, amputees using passive knee joints often demonstrate a slower than normal swing phase [39]. A significant asymmetry in either stance time or swing time (they are correlated by cadence) is often indicative of non-optimal gait. In our particular control strategy, the user has direct control over the leg during stance and relatively less control over the prosthesis during swing. . If the prosthesis is taking slower or longer than a nominal duration during swing, than it also forces asymmetry during the stance times. Hence, the first goal is to ensure that swing times for both prosthetic and sound sides are close together, though certain amount of asymmetry is available due to the nature of the testing adapter.

KINEMATIC MEASURE

Joint trajectory data and certain kinematic markers can be used to assess gait patterns. However, before we begin to discuss the relevant features, it is to be acknowledged that fundamental mechanical distinctions between the prosthesis and a real limb together with the fact that user is fitted with a brace, limit the extent to which natural looking data can be replicated in this test setup. Nonetheless, we identify certain kinematic pointers that provide some measure of improvement in joint patterns and are listed in Table 9-1. We are primarily interested in two aspects of kinematic parameters, (1) they occur in the right percent of stride range and (2) values are within reasonable range of biomechanical data. Some particular measures such as stance flexion are particularly hard to reproduce because the adapter brace partially slides up the hip during heel strike absorbing impact and lowering the body, thus no flexion at the knee is necessary. However, the values of the rest of the parameters are intuitively tuned from recorded data and the feedback of the user. Noticeable deviations from range of biomechanical data are avoided by further tuning the gait variables.

Table 9-1. Kinematic variables useful for the evaluation of gait

Mode	Kinematic Measures	Range of Values	Percent of Stride
Stance Flexion – Extension	(1) Max Knee Stance Flexion (2) Max Ankle Flexion (3) Percent of stride when ankle angle becomes flexive	(1) 15 – 25 deg (2) 7 – 10 deg	(1) 12 – 15% (2) 41 – 45% (3) 5 – 8%
Pre-Swing / Push off	(1) Ratio of Knee to Ankle angle	(1) 1.2 – 1.4	
Swing Flexion	(1) Max Swing Flexion Angle	(1) 60 – 70%	(1) 69 – 73%
Swing Extension	(1) Knee angle at end of forward swing	(1) 1 – 3 deg	(1) 96 – 98%

Another, useful variable in the assessment of gait is the measurement the joint torque profiles. Noticeable lag of desired max torques in the angle for instance, indicate insufficient power during push off phase. In addition, higher than normal knee torques at the joint during push off can indicate unwanted stiffness in knee. Thus, cycle-to-cycle joint torques is measured to monitor for product of torques in desired range of values and to ensure that switching between modes does create a large step in torques.

TUNING

The tuning of the parameters is done in an intuitive manner through user feedback and observation of gait parameter. For instance if the user felt the knee to be too weak during stance, then stiffness would be increased. Similarly, if the ankle during push off was too weak or did not act fast enough, then the stiffness and set point changes would be appropriately adjusted. Timing information and recorded kinematic variables were used to monitor whether sufficient torques were begin generated, and whether gait transitions happened at correct times. In addition, important kinematic variables (Table 9-1) and kinetic variables provided clues to sources of errors. However, since the user is fitted with an able bodied adapter that alters the dynamics and introduces asymmetry, the question of when an optimal gait pattern is reached is difficult to determine. The gait trials were performed for two speeds (1.5 and 1.8 mph), and both yielded similar gait patterns after tuning. Hence, once the user after successive tuning felt comfortable walking at a given speed, and consecutive adjustments of impedance parameters did not produce any noticeable variation in the gait patterns, no further tuning was performed.

CHAPTER X

EXPERIMENTAL RESULTS AND DISCUSSION

The results of the gait trial performed on the treadmill at a walking speed of 0.8 m/s (1.8mph) are reported here. As mentioned earlier the results presented here are those obtained through iterative tuning of the leg, until no further noticeable improvement could be seen in the gait patterns or felt by the user. Though, walking trails were also conducted at a slower speed of 1.5 mph, the gait patterns do not show distinct differences and the tuning parameters were very similar. In addition, the gait patterns were also significant dependent on the individuals walking pattern. Gait pattern and the users response would remain good during the initial to middle part of a session that would last about an hour. As the user fatigued the gait patterns were likely to undergo gradual deterioration and cycle to cycle repeatability would be affected. The time periods, kinematic, kinetic and power patterns of the gait shown here are representative of those collected from the middle time period of a session (i.e. after the user has acclimatized and before fatigue becomes a factor)

The first response of a user walking with prototype is a decreased cadence (i.e. increase stride period) with an increasing stride length, which is often typical of amputees. The average stride period was found to be around 1.56 seconds, which is noticeably less than the user walking freely on the treadmill. As the user walks with an increases stride period, the time spent on each leg is uneven. The user generally spends longer time on his sound leg during stance as indicated by the longer swing period of prosthetic leg (Fig 10-1). Though, the average difference in the swing time period are about 100 ms, they are representative of much improved swing times. Swing times prior to final tuning display differences of more that 200 ms, and gait is visibly asymmetrical. The swing times were improved upon by increasing the stiffness component of the knee joint during swing extension to influence a faster forward swing. However, continuously strengthening the stiffness did not result in increased swing times, since the user would hold the knee at full extension for a while before heel strike. Thus, past some threshold period, decreasing the forward swing period made the user vulnerable to stumbles and unnatural looking swing behavior, while providing no significant improvement in temporal asymmetry. The sound side leg swing time can also undergo variation as can be seen towards the latter end of the graph (Fig 10-1) as is most likely a response to a non-ideal stance behavior on the prosthetic side. Ultimately the asymmetry in swing times is likely due to the distinctly different inertial properties of the prosthesis and natural limb, thus a certain amount of difference in the time period is to be expected.

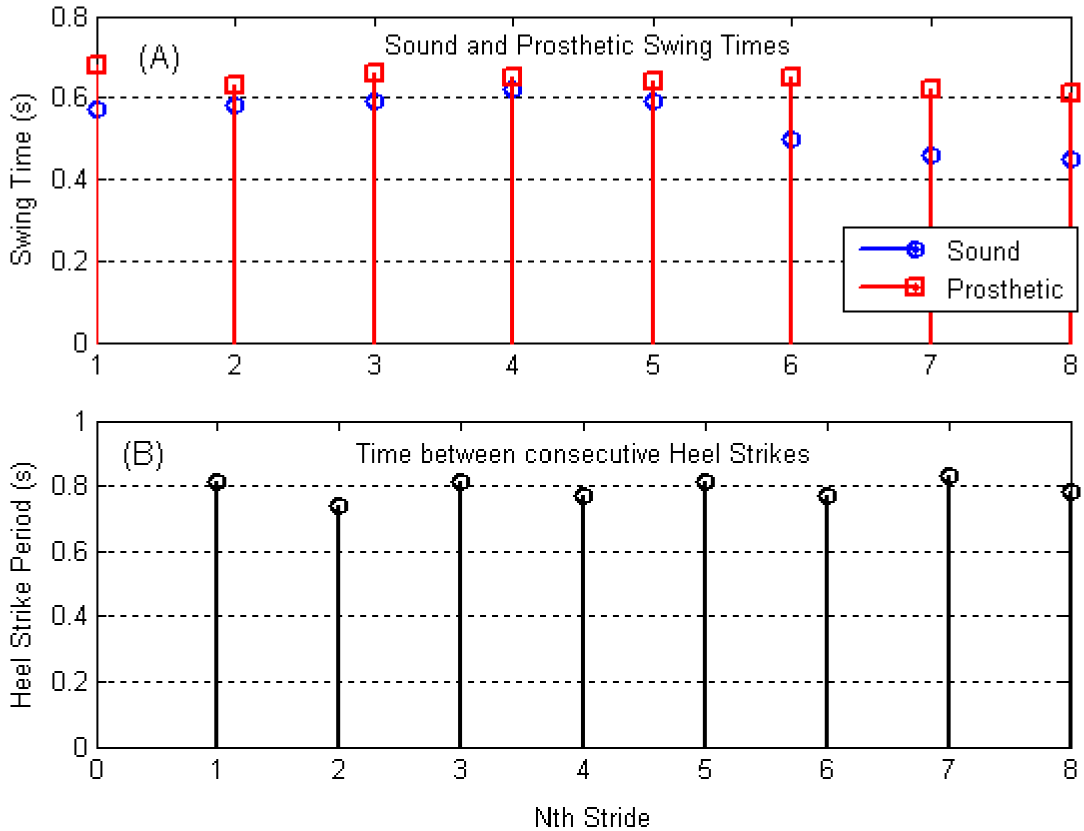


Figure 10-1. (A) Average swing time for prosthetic and sound leg for 8 consecutive stride cycle. The prosthetic exhibits longer swing times than the sound side. (B) Time period between consecutive 8 heel strikes. The longer swing times for the prosthetic also influence the uneven heel strikes period

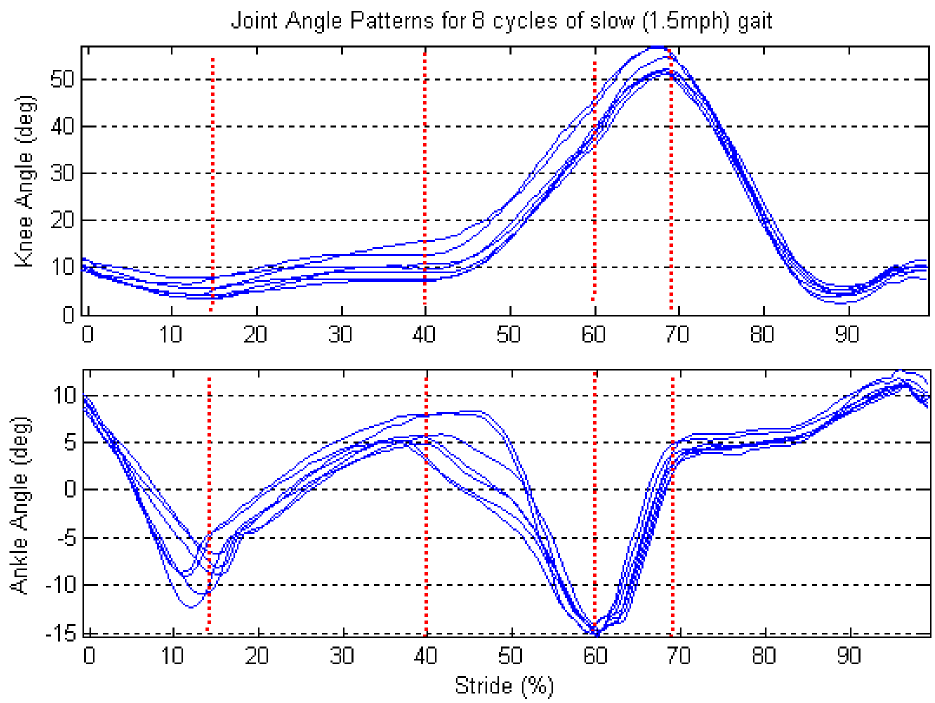


Figure 10-2. Measured joint angles (degrees) for six consecutive gait cycles for a treadmill walk (1.8mph). Average stride period is 1.56s. The red lines indicates mode switching as outlined in the finite state gait model (Fig 8-1)

The joint angles for 8 consecutive cycles of gait are show in Fig 10-2. The joint patterns show very good similarity to gait patterns of normal walking (Fig 3-3) except during stance flexion (0 – 40 %). Most noticeable is the absence of stance flexion in knee joint after heel strike. The difference in behavior during this period is most likely a result of the significant compliance between the adaptor and user. Specifically, the role of the knee during this period is to flex slightly upon impact, which absorbs energy and cushions the impact of heel strike. As such, the knee acts effectively as a stiff spring, first absorbing the energy of impact and shortly after, returning this energy to the user. When used with the adaptor, this knee stiffness acts in series with the (much lower) stiffness of the user/adaptor interface, and thus the cushioning role of knee flexion during heel strike is dominated by the compliance in the user/adaptor interface. This behavior is evident in the video recordings of the gait trials by watching the relative motion between the top of the brace and the subject's hip during heel strike. Once the axial compliance between the user and prosthesis is reduced significantly (as would be the case with an amputee subject), the knee joint is likely to exhibit the flexion and subsequent extension evidenced in the prototypical gait kinematics of Fig. 3-6.

Beside stance flexion-extension, the joint patterns are reasonably close to normal biomechanical patterns. The swing flexion is slightly lower than normal values; however this was to be expected due to issues of back drivability in the actuators and did not interfere significantly with gait. The ankle flexed past 5 degrees prior to undergoing an impulsive extension to provide push off.

A relatively harder task was to tune the knee and ankle joint to coordinate toe-off. During Toe off, the ankle goes through rapid extension to provide an impulsive push, while the knee simultaneously flexes to avoid any jerks in the body due to the extension of the ankle. It was found to be particularly beneficial to tune the ankle first during this phase, to ensure proper interaction with the ground to generate necessary torques. Once, the ankle was tuned to generate the desired ankle torque with in a set amount of range of ankle extension, the knee stiffness was then tuned to enable a certain amount of correlated flexion in the knee. The standard ratio of knee to ankle angle during push off was found to be between 1.2 – 1.4 through population data [1], and the measured average ratio for the trial was 1.2. The ratio of knee to ankle angle was found to be a useful measure of position interaction between the joints during toe off.

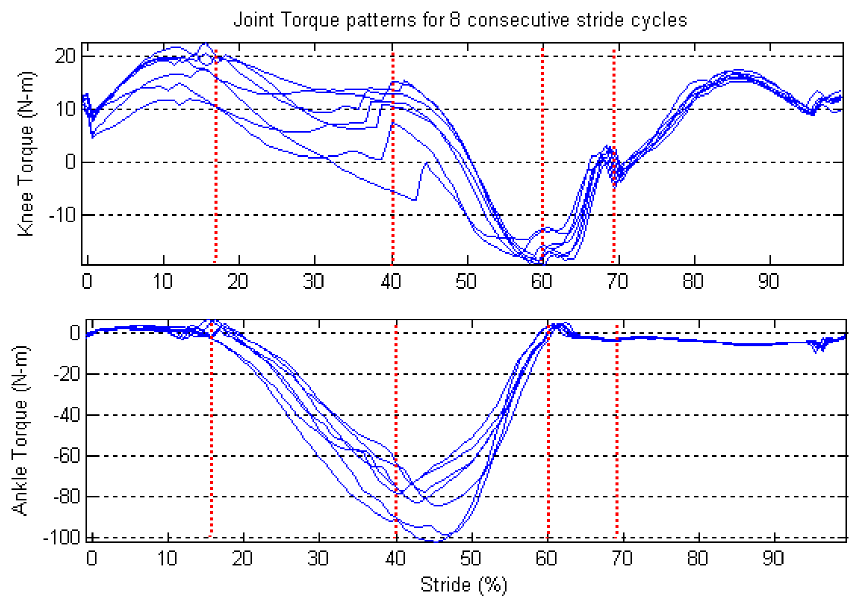


Figure 10-3. Measured joint torques (N.m) for eight consecutive gait cycles for a treadmill walk (1.8mph). Average stride period is 1.56s. The red lines indicate mode switching as outlined in the finite state gait model (Fig 23)

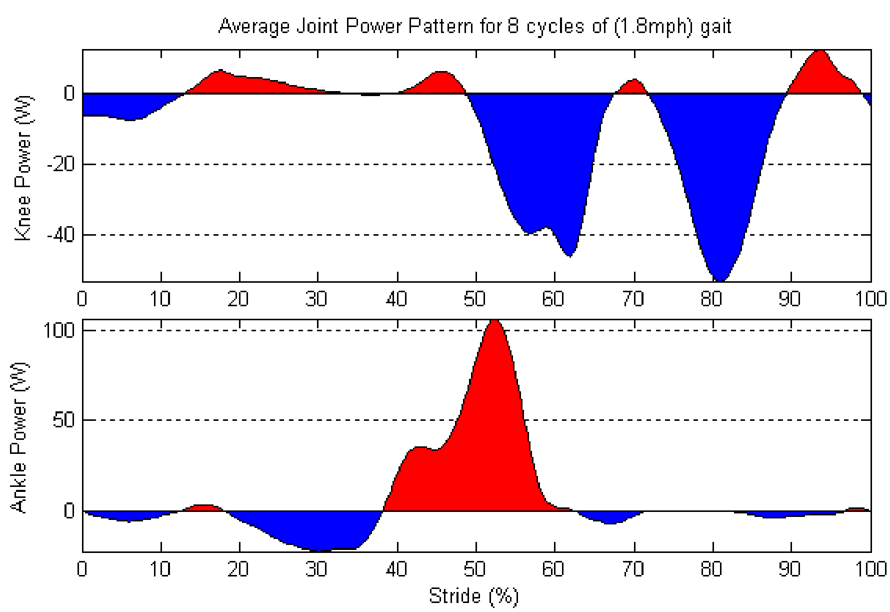


Figure 10-4. Average Joint Power for eight consecutive gait cycles for a treadmill walk (1.8mph). Average stride period is 1.56s. Red indicates positive power output and blue the negative power dissipation

The measured joint torques (Fig 10-3) during the gait show high cycle-to-cycle variability during the stance flexion mode (0 – 40 %), which as discussed earlier is likely due to the compliant interaction between the adapter brace and the user. The torque levels for the rest of the modes for both knee and ankle compare quite well to average population data (Fig 5-1). Sharp transition due to certain transitions behavior can be seen in the torque profile, however, they are too small to noticeably affect the user. In fact, the tuning of parameters is also done in such a manner that the transition between modes do not result in significantly high ‘step’ in forces.

The knee and ankle joint powers, which were computed directly from the torque and differentiated angle data, are shown in Fig. 10-4, and indicate that the prosthesis is supplying a significant amount of power to the user. As discussed earlier, a small positive push is provided during swing (70% of stride) to improve symmetry of swing times between the sound and prosthetic side. The large damping behavior of the knee joint during swing is also very apparent. However, the noticeable feature is the large power output at the ankle joint during push off. Note that the measured power compares favorably to that measured for healthy subjects (Fig 1-1), and thus indicates an enhanced level of functionality relative to existing passive prostheses

The final set of tuned impedance parameters for the knee and ankle joint in each mode are listed in Table 10-1.

Table 10-1. Impedance parameters derived by experimental tuning

Mode	**Knee Impedance**			**Ankle Impedance**		
	k_1 (N.m/deg))	b (N.m.s)	θ_e (deg)	k_1 (N.m/deg))	b (N.m.s)	θ_e (deg)
0	2	0	13	.04	0	-1
1	1.25	0	18	3	0	-4.4*
2	1.0	0.	16	4.5	0	-15
3	0.5	0.01	55	0.32	0	0
4	0.4	0.02	35	0.32	0	0

* Ankle Threshold = -3.5 degrees, λ = 2.1

CHAPTER XI

CONCLUSIONS

A number of opportunities and challenges lie ahead towards the realization of functional powered transfemoral prosthesis. This project has explored a possible avenue of an impedance based control framework that allows for the generation of torques at the prosthetic joints in a stable manner. Gait trials with an able bodied adapter was validated against population gait data and the method has demonstrated sufficient strength to further explore its feasibility. However, further improvement of the control framework will have to go hand in hand with refinements of hardware and test setup. The able-bodied adapter provides a useful setup for preliminary testing of the control architecture, though further enhancements will need to be made to achieve greater realism in simulating normal gait. Higher output impedance at the joint due to non-back drivability of the pneumatic actuator also presents a significant challenge, which could partly be addressed by improving upon the valve mechanism. The overall control could benefit from improved valve bandwidth and reduced lag in the air supply; and changes are already underway to address these issues.

Regarding the controller, several further improvements are necessary. A reliable quantitative method of evaluating gait with an improved sensor set is the foremost of challenges and one that could significantly improve the ability to deliver better results. Moving towards partial or full automation of tuning of impedance model parameters could significantly reduce rehabilitation and testing times. This is likely to require a broader understanding of the variation in parameters with different gait speed. The impedance model could be further expanded to cover steps and ramp walking to improve functionality. Ultimately, the controller is likely to be used in conjunction with an intelligent user intent recognition system to deliver a complete set of locomotive functionality to an amputee.

Finally, the incredible versatility that human show in adapting and coordinating with external devices is demonstrated. If the prosthesis are designed to suitably approximate normal limb dynamics with sufficient power, than humans have the capability to leverage it effectively to aid locomotion. We hope to further pursue the design and control frame work for powered prosthetic technology through this perspective

APPENDIX A: CONSTRAINED PIECEWISE FITTING OF KNEE AND ANKLE TORQUES DURING NORMAL SPEED LEVEL WALK

Matlab code for “Constrained Piecewise fitting of knee and ankle torques during normal speed level walk” to data Provided by Winter [1]

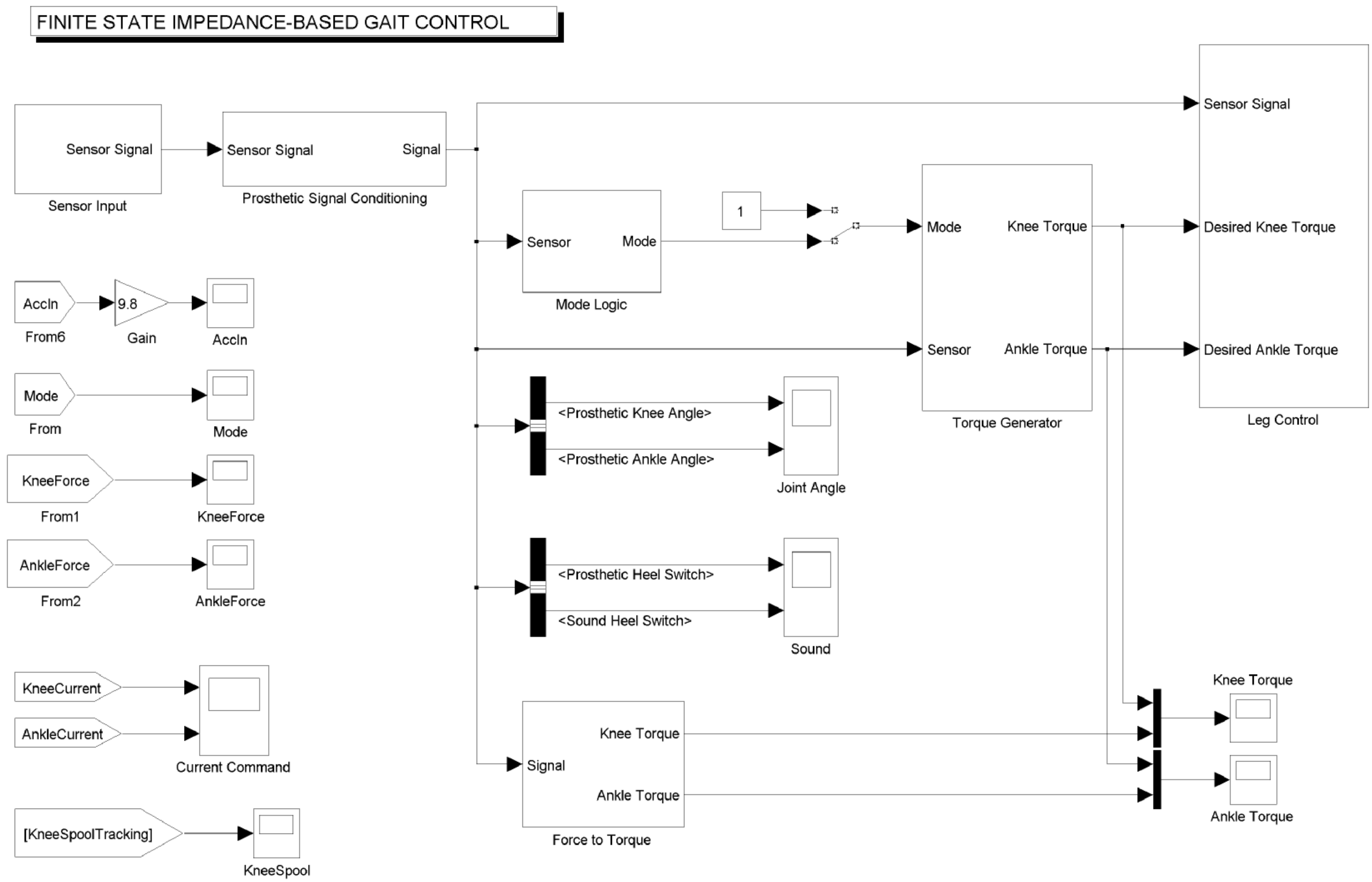
```
%Piecewise constrained fitting of gait data to impedance model
%(Linear Spring + Damper)

clear

%Load desired Winter, cadence data
load('../../Winter Data/normal_cadence_data.mat');

%Estimated mass of user to use with normalized data
mass = 75;

%Display splined Winter reported knee and ankle torque,
%for a complete stride cycle

figure
m = mass;
subplot(2,1,1);
xx = 0:0.1:100;
yy = spline(stride,-m.*knee_torque,xx);
plot(xx,yy,'LineWidth',2);
hold on

subplot(2,1,2);
yy = spline(stride,-m.*ankle_torque,xx);
plot(xx,yy,'LineWidth',2);
hold on;

%Specify Partition of a stride cycle into four modes
%range1 = 1:19;range2 = 20:32; range3 = 33:37; range4 = 38:51;
%The following specifies the array index for the desired
%paritition range in pairs, eg. the following partitions the
%gait into 0-42%, 44-66%, and so on
ranges = [1 22 23 31 32 37 38 51];

%Temporarily set mass to '1' so as to use the Winter nomalized data
%directly
m = 1;

%Array to store parameter and output of constrained LS fit
KneefitYY = []; Kneeparams = [];
AnklefitYY = []; Ankleparams = [];
```

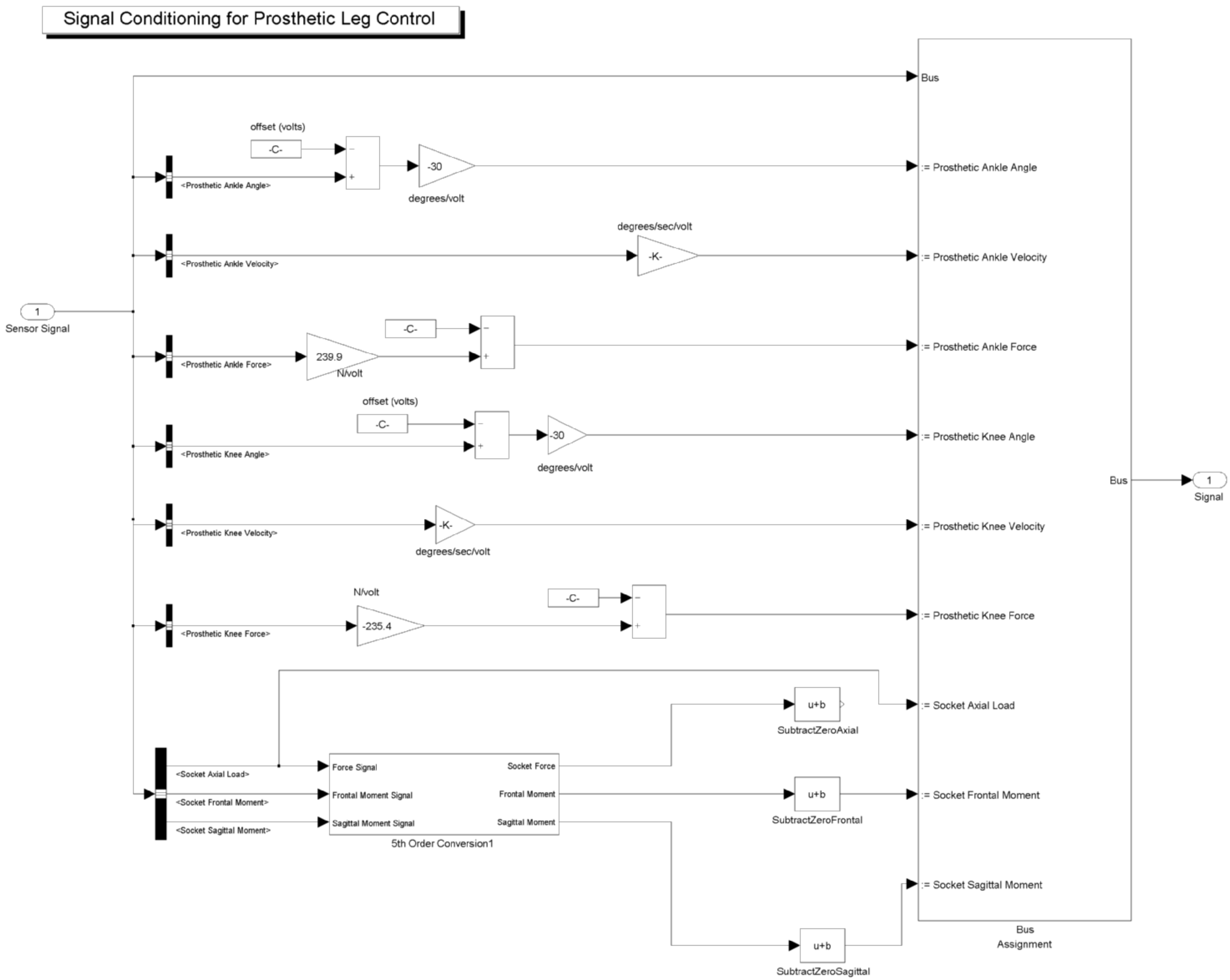
```
%Array to store ratio of stiffness to damping terms predicted by fit
KneeRatio = []; AnkleRatio = [];

%Store equillibrium angle for knee and ankle spring
KneeTheta = 0; AnkleTheta = 0;
k=1;
%Iterate through each partition specified in the range...
for (i = 1:2:length(ranges))

   %Current range
   range = (ranges(i)):(ranges(i+1));

    %For each iteration find a suitable equilibrium point, that
    minimizes
   %the standard deviation measure.
   KstdMeasure = 100; AstdMeasure = 100;
   %Specify settings for the constrained fitting.
   opt = optimset('Display','off','TolFun',1e-8,'TolX',1e-7);

    %Iterate through a reasonable range of theta to find an optimal
    value
   for (theta = -50:1:100)

      %%%%%Fit Knee torques%%%%%%%%%
      %Setup X & Y variables for LS fitting for knee
      KXX = -1.*[(knee_angle(range)-theta) knee_vel(range)];
      KYY = (-m.*knee_torque(range));

      %Constrained non-negative Least Squares (LS) fitting
      kb = lsqnonneg(KXX,KYY,1e-3.*[5;0],opt);

      %%%%%Fit Ankle torques%%%%%%%%%
      %Setup X & Y variables for LS fitting for ankle
      AXX = -1.*[(ankle_angle(range)-theta) ankle_vel(range)];
      AYY = (-m.*ankle_torque(range));

      %Constrained non-negative Least Squares (LS) fitting
      ab = lsqnonneg(AXX,AYY,1e-3.*[5;0],opt);

      %Calculate the Mean Square of the error residual
      AnkleFitError = sum((AYY-AXX*ab).^2);
      KneeFitError = sum((KYY-KXX*kb).^2);

      %Compare if this choice of equillibrium theta provides a smaller
      %error residual, if it does store it as the optimal theta
      if (KneeFitError < KstdMeasure)
         KstdMeasure = KneeFitError;
         KneeTheta = theta;
      end
      if (AnkleFitError < AstdMeasure)
         AstdMeasure = AnkleFitError;
         AnkleTheta = theta;
      end
   end   %End search for optimal equilibrium theta
```

```
    %Use the optimal equillibrium angle to performed LS fit
    KX1 = -1.*[(knee_angle(range)-KneeTheta) knee_vel(range)];
    AX1 = -1.*[(ankle_angle(range)-AnkleTheta) ankle_vel(range)];
    KY1 = [-m.*knee_torque(range)];
    AY1 = [-m.*ankle_torque(range)];

    %Find best fit using constrained least squares
    %Fitting is sensitive to choice of initial conditions
    kb1 = lsqnonneg(KX1,KY1,1e-3.*[5;0],opt);
    ab1 = lsqnonneg(AX1,AY1,1e-3.*[5;0],opt);

    %Determining the contribution of stiffness and damping terms

    %For Knee
    KY = KX1*kb1;
    kd_contbn = kb1(2).*knee_vel(range);   %Damping contribution
    ks_contbn = KY-kd_contbn;             %Stiffness contribution
    %Ratio of stiffness to damping contribution
    k_ratio(k) = mean(abs(ks_contbn./kd_contbn));

    %For Ankle
    AY = AX1*ab1;
    ad_contbn = ab1(2).*ankle_vel(range);   %Damping contribution
    as_contbn = AY-ad_contbn;             %Stiffness contribution
    %Ratio of stiffness to damping contribution
    a_ratio(k) = mean(abs(as_contbn./ad_contbn));

    %Store output predicted from constrained LS fitting
    KneefitYY = [KneefitYY; KX1*kb1];
    AnklefitYY = [AnklefitYY; AX1*ab1];

    %Store Impedance parameters for this stride partition
    kb1(4) = KneeTheta;
    Kneeparams = [Kneeparams kb1];
    ab1(4) = AnkleTheta;
    Ankleparams = [Ankleparams ab1];

    %Plot Vertical Lines indicating partition boundary of stride
    m = mass;
    subplot(2,1,1);
    %plot(stride(range),m.*KX1*kb1(1:3),'--g','LineWidth',3);
    plot([stride(range(end)) stride(range(end))],m.*[-1
0.5],'--r','LineWidth',1)
    %text(stride(range(3)),-.75,num2str(k_ratio(k),3));
    subplot(2,1,2);
    hold on
    %plot(stride(range),m.*AX1*ab1(1:3),'--g','LineWidth',3)
    plot([stride(range(end)) stride(range(end))],m.*[-2
1],'--r','LineWidth',1)
    %text(stride(range(3)),-1.5,num2str(a_ratio(k),3))
    m =  1;
    k=k+1;
end
```

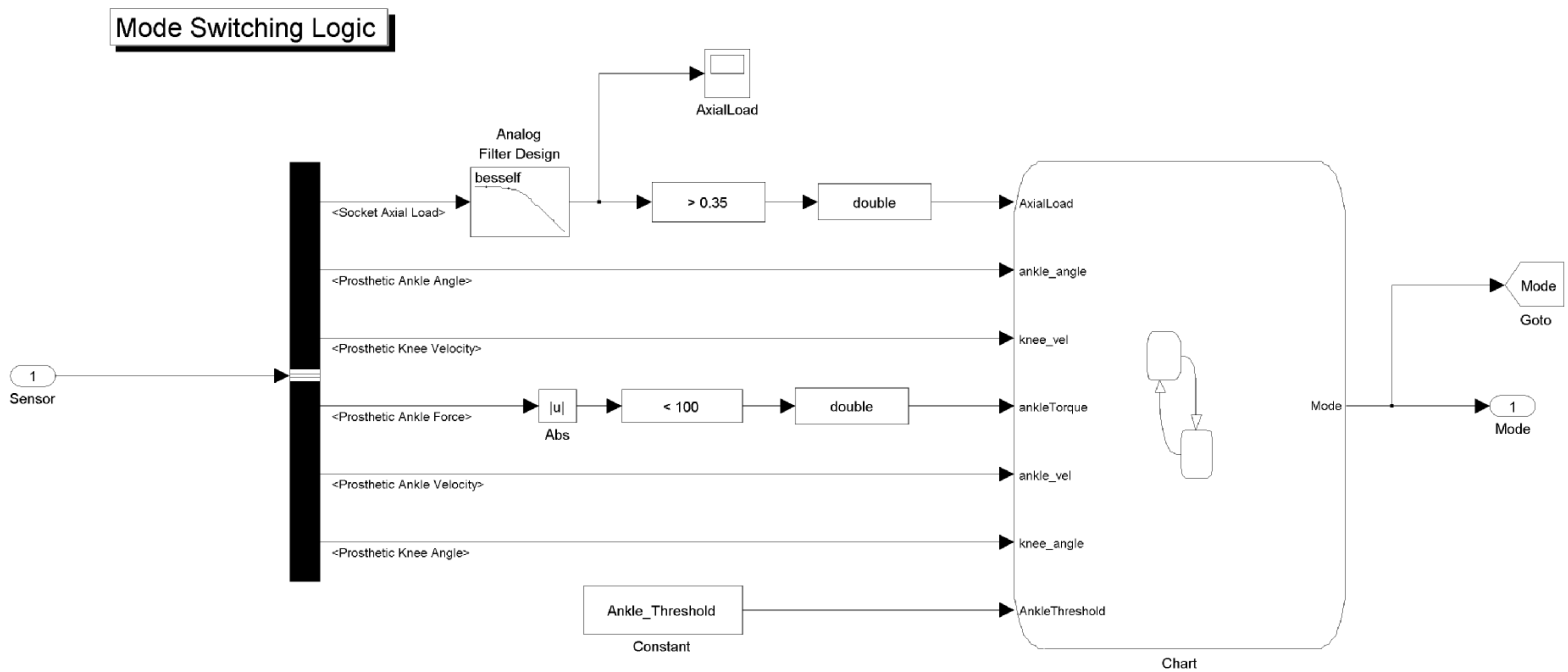
```
m = mass;
%To avoid sharp jumps at the transition boundaries,
%the entire output of the fitted model is splined together
%and plotted
subplot(2,1,1);
title('Impedance fit of Knee and Ankle Joint : Linear spring + Damper');
X = stride(ranges(1):ranges(end));
Y = KneefitYY.*m;
xx = 0:1:100;
yy = spline(X,Y,xx);
plot(xx,yy,':r','LineWidth',3);
ylabel('Joint Torque');

subplot(2,1,2);
X = stride(ranges(1):ranges(end));
Y = AnklefitYY.*m;
xx = 0:1:100;
yy = spline(X,Y,xx);
plot(xx,yy,':r','LineWidth',3);
xlabel('Percent of Stride');

disp('KNEE');
disp(['      K1       K2       b        ThetaO'])
disp(Kneeparams');
disp('ANKLE');
disp(['      K1       K2       b        ThetaO'])
disp(Ankleparams');
```

APPENDIX B.SIMULINK IMPLEMENTATION OF THE CONTROLLER TOGETHER WITH THE FINITE STATE MODEL

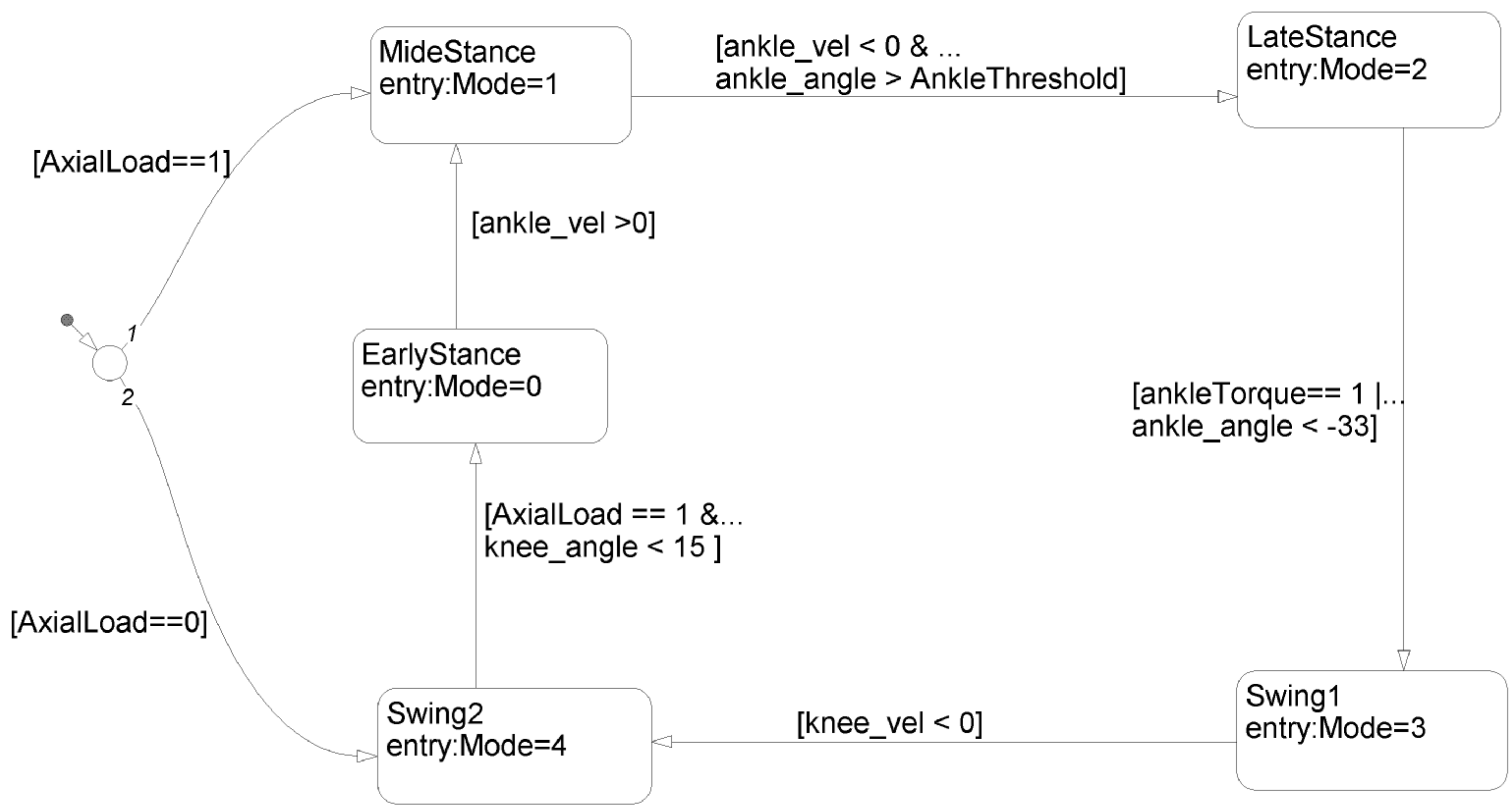


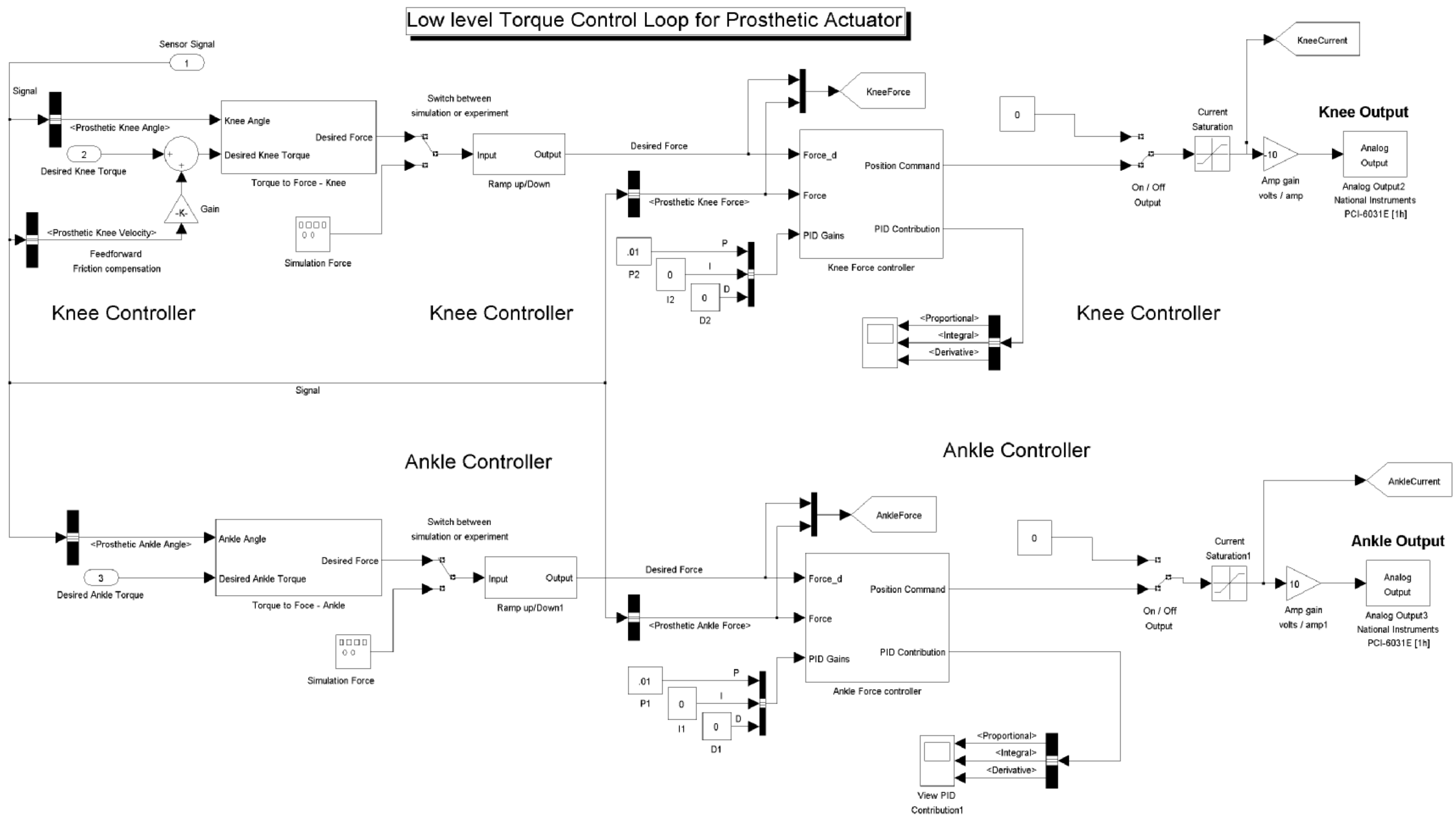


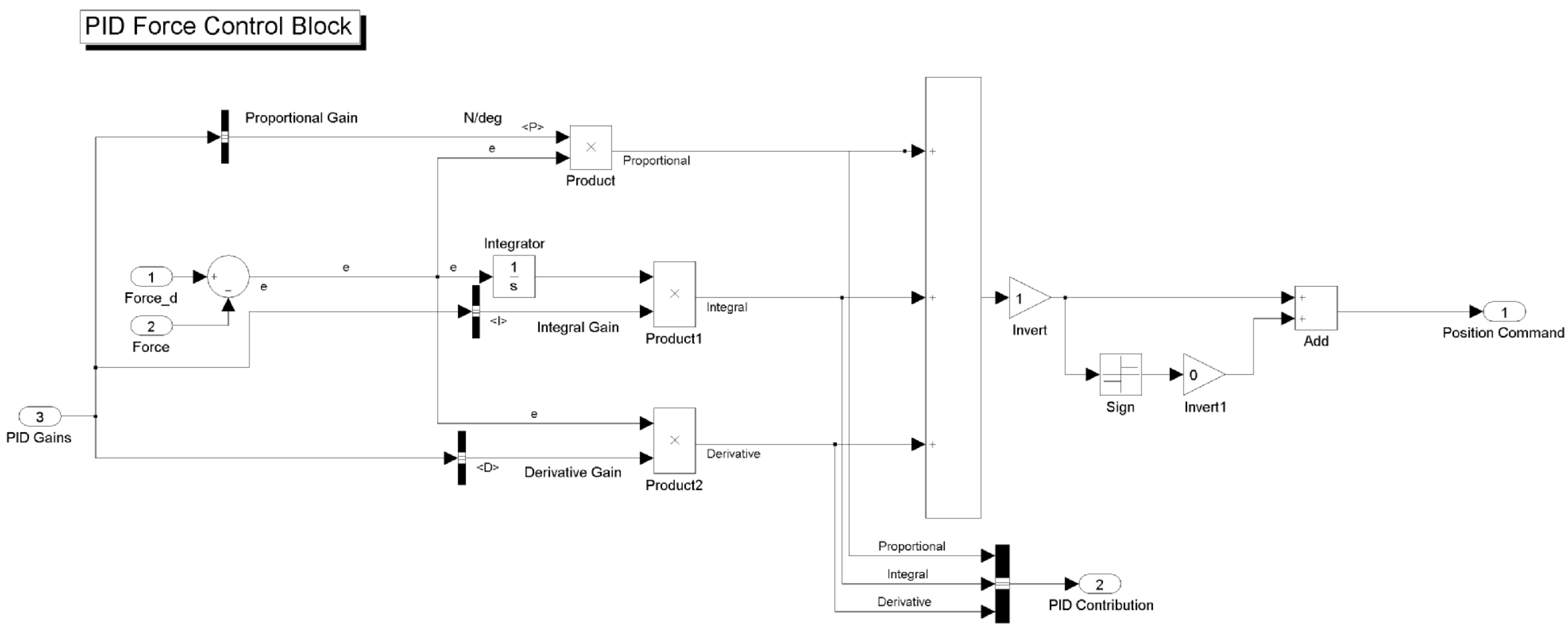
Signal Conditioning for Prosthetic Leg Control
1
Sensor Signal
offset (volts)
-C-
<Prosthetic Ankle Angle>
-30
degrees/volt
degrees/sec/volt
-K-
<Prosthetic Ankle Velocity>
-C-
<Prosthetic Ankle Force>
239.9
N/volt
offset (volts)
-C-
<Prosthetic Knee Angle>
-30
degrees/volt
<Prosthetic Knee Velocity>
-K-
degrees/sec/volt
N/volt
<Prosthetic Knee Force>
-235.4
-C-
<Socket Axial Load>
<Socket Frontal Moment>
<Socket Sagittal Moment>
Force Signal
Frontal Moment Signal
Sagittal Moment Signal
Socket Force
Frontal Moment
Sagittal Moment
5th Order Conversion1
u+b
SubtractZeroAxial
u+b
SubtractZeroFrontal
u+b
SubtractZeroSagittal
Bus
:= Prosthetic Ankle Angle
:= Prosthetic Ankle Velocity
:= Prosthetic Ankle Force
:= Prosthetic Knee Angle
:= Prosthetic Knee Velocity
:= Prosthetic Knee Force
:= Socket Axial Load
:= Socket Frontal Moment
:= Socket Sagittal Moment
Bus Assignment
Bus
1
Signal


Mode Switching Logic
1
Sensor
<Socket Axial Load>
<Prosthetic Ankle Angle>
<Prosthetic Knee Velocity>
<Prosthetic Ankle Force>
<Prosthetic Ankle Velocity>
<Prosthetic Knee Angle>
Analog
Filter Design
besself
AxialLoad
> 0.35
double
|u|
Abs
< 100
double
Ankle_Threshold
Constant
AxialLoad
ankle_angle
knee_vel
ankleTorque
ankle_vel
knee_angle
AnkleThreshold
Mode
Chart
Mode
Goto
1
Mode

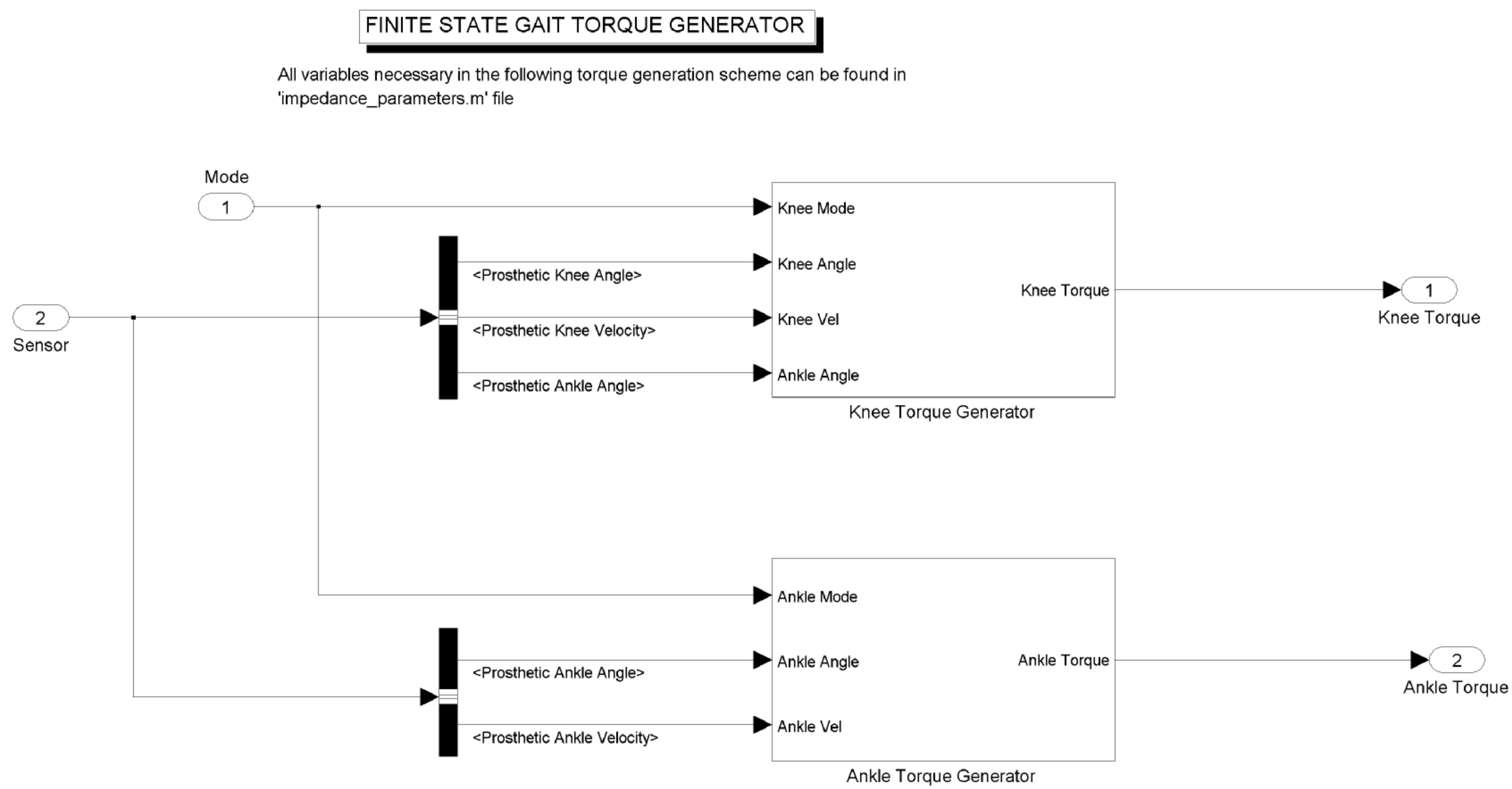
FINITE STATE MODEL OF GAIT IN STATEFLOW

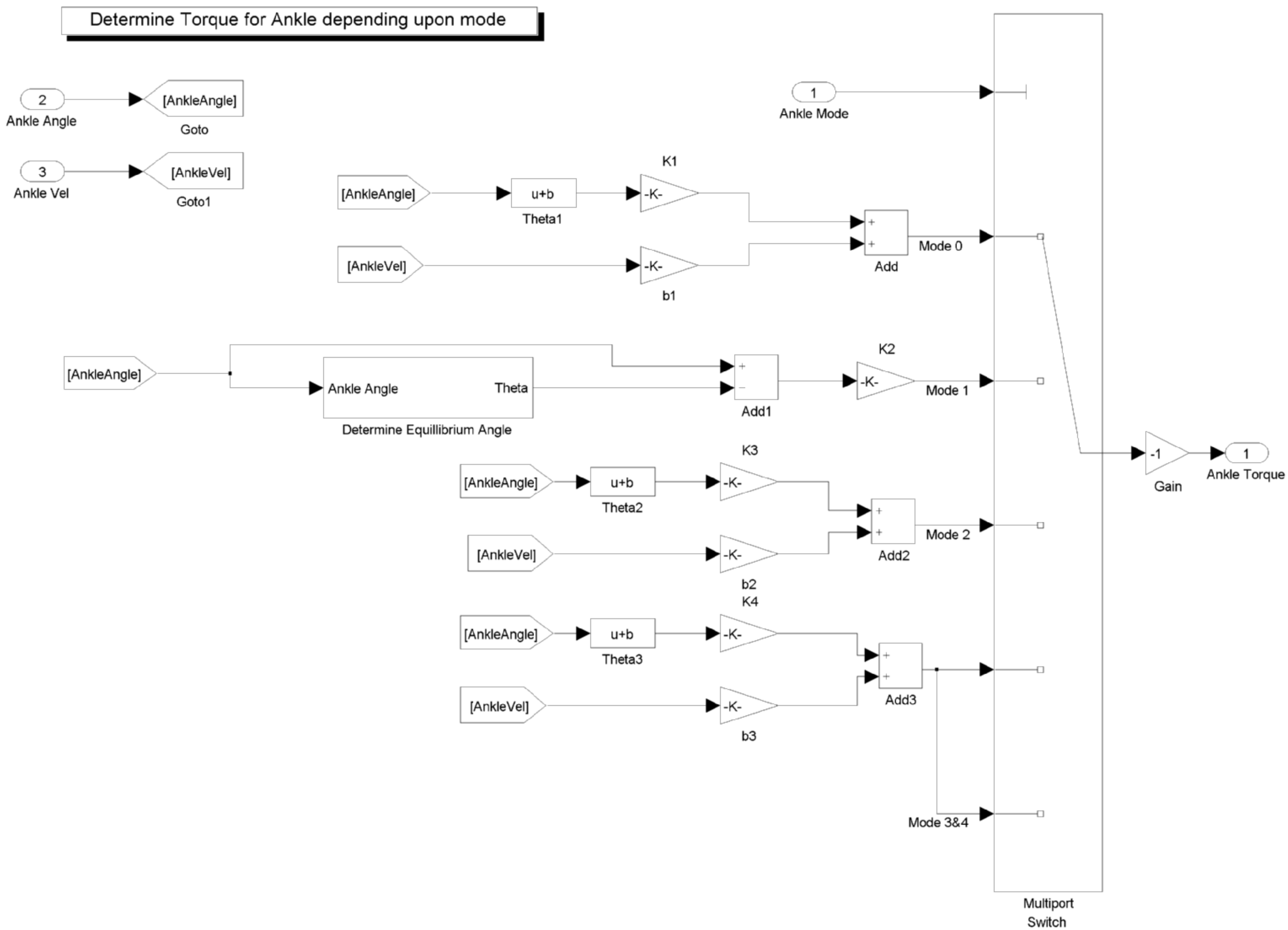



Low level Torque Control Loop for Prosthetic Actuator
Sensor Signal
1
Signal
<Prosthetic Knee Angle>
2
Desired Knee Torque
Gain
-K-
<Prosthetic Knee Velocity>
Feedforward
Friction compensation
Knee Angle
Desired Knee Torque
Desired Force
Torque to Force - Knee
Simulation Force
Switch between
simulation or experiment
Input
Output
Ramp up/Down
Desired Force
<Prosthetic Knee Force>
KneeForce
Force_d
Force
PID Gains
Position Command
PID Contribution
Knee Force controller
.01
P2
0
I2
0
D2
P
I
D
<Proportional>
<Integral>
<Derivative>
0
On / Off
Output
Current
Saturation
KneeCurrent
10
Amp gain
volts / amp
Knee Output
Analog
Output
Analog Output2
National Instruments
PCI-6031E [1h]
Knee Controller
Knee Controller
Knee Controller
Signal
Ankle Controller
Ankle Controller
<Prosthetic Ankle Angle>
3
Desired Ankle Torque
Ankle Angle
Desired Ankle Torque
Desired Force
Torque to Foce - Ankle
Simulation Force
Switch between
simulation or experiment
Input
Output
Ramp up/Down1
Desired Force
<Prosthetic Ankle Force>
AnkleForce
Force_d
Force
PID Gains
Position Command
PID Contribution
Ankle Force controller
.01
P1
0
I1
0
D1
P
I
D
<Proportional>
<Integral>
<Derivative>
View PID
Contribution1
0
On / Off
Output
Current
Saturation1
AnkleCurrent
10
Amp gain
volts / amp1
Ankle Output
Analog
Output
Analog Output3
National Instruments
PCI-6031E [1h]



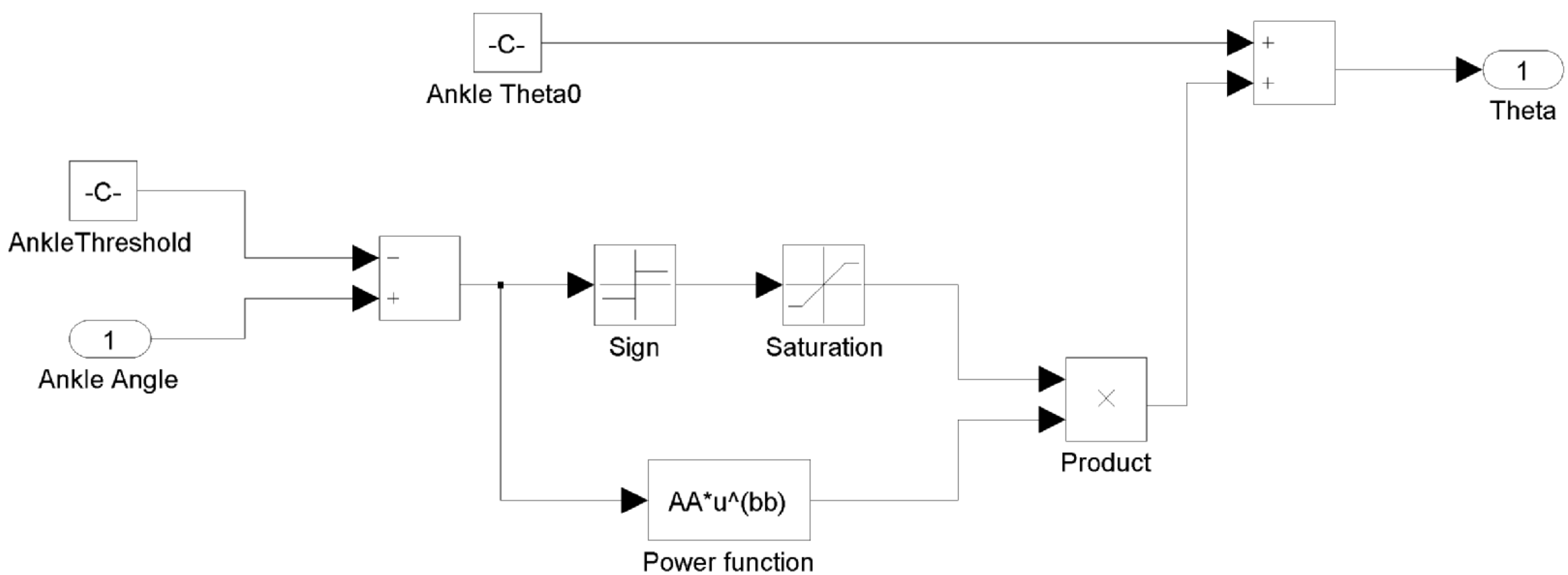
PID Force Control Block
Proportional Gain
N/deg
<P>
e
Product
Proportional
Integrator
1
s
e
e
e
1
Force_d
2
Force
<I>
Integral Gain
Product1
Integral
3
PID Gains
e
<D>
Derivative Gain
Product2
Derivative
1
Invert
Sign
0
Invert1
Add
1
Position Command
Proportional
Integral
Derivative
2
PID Contribution

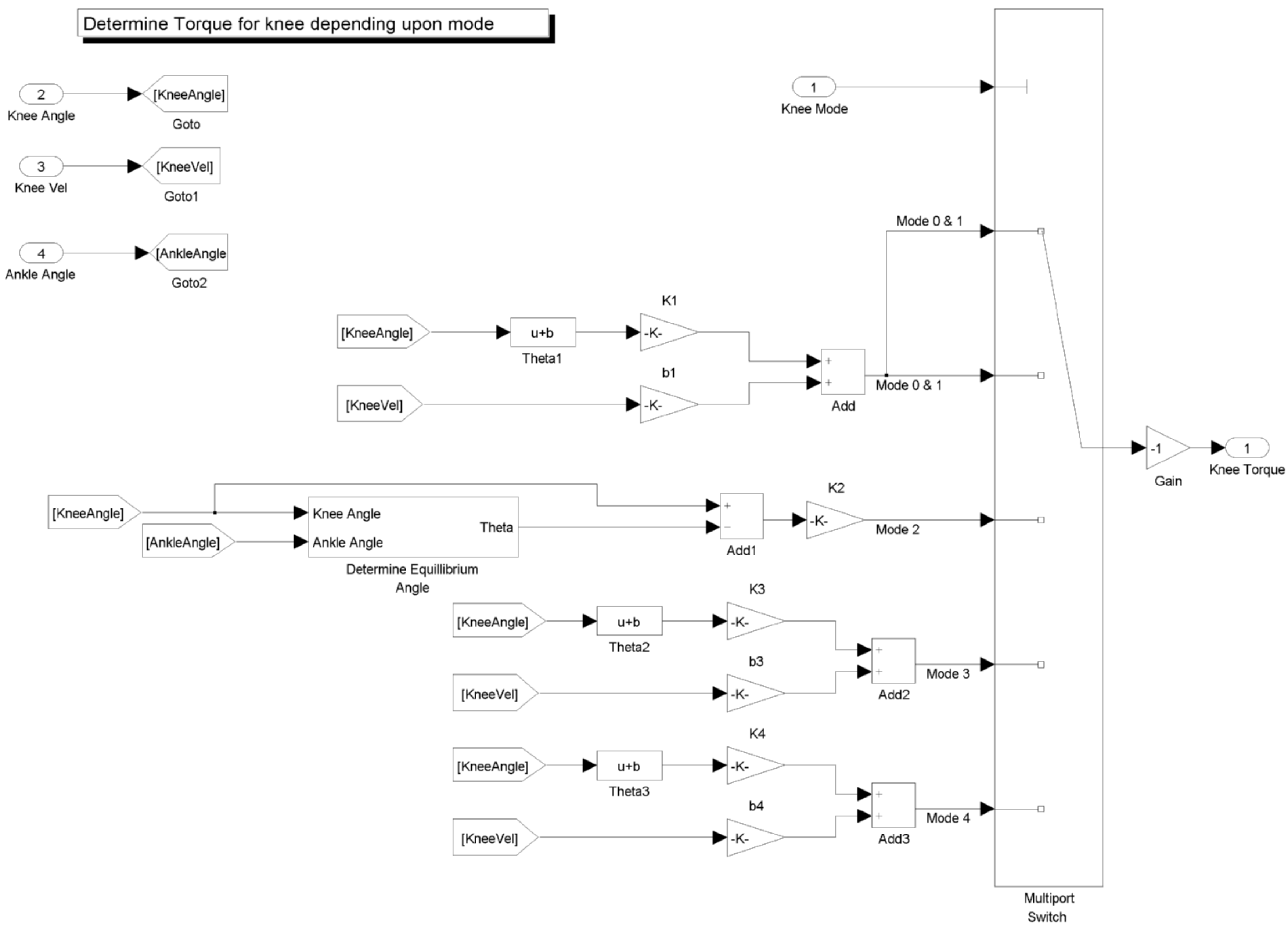

FINITE STATE GAIT TORQUE GENERATOR
All variables necessary in the following torque generation scheme can be found in 'impedance_parameters.m' file
Mode
1
2
Sensor
<Prosthetic Knee Angle>
<Prosthetic Knee Velocity>
<Prosthetic Ankle Angle>
Knee Mode
Knee Angle
Knee Vel
Ankle Angle
Knee Torque
Knee Torque Generator
1
Knee Torque
<Prosthetic Ankle Angle>
<Prosthetic Ankle Velocity>
Ankle Mode
Ankle Angle
Ankle Vel
Ankle Torque
Ankle Torque Generator
2
Ankle Torque



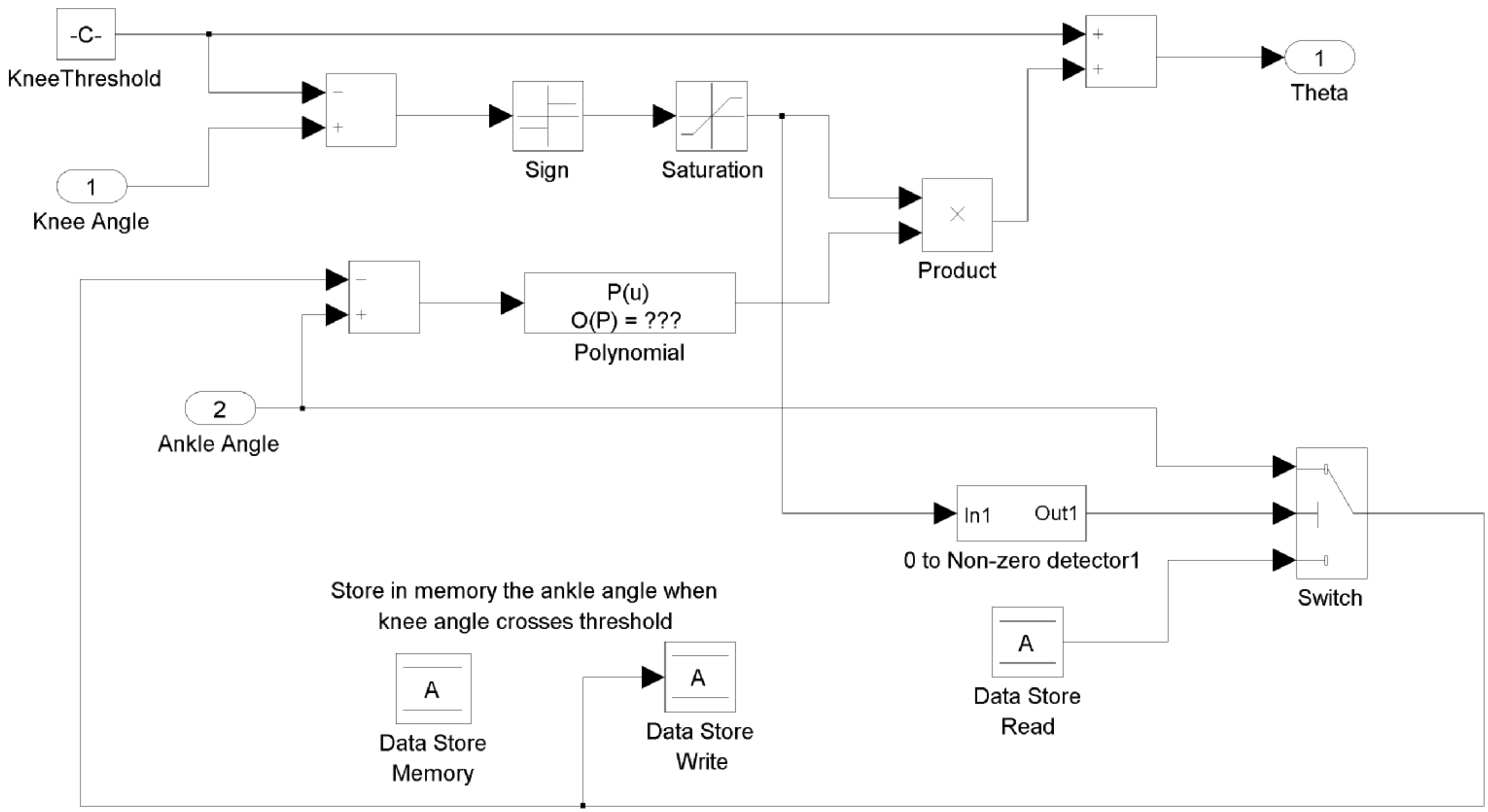
Determine Torque for Ankle depending upon mode
2
Ankle Angle
[AnkleAngle]
Goto
3
Ankle Vel
[AnkleVel]
Goto1
1
Ankle Mode
[AnkleAngle]
u+b
Theta1
K1
-K-
[AnkleVel]
-K-
b1
Add
Mode 0
[AnkleAngle]
Ankle Angle
Theta
Determine Equillibrium Angle
Add1
K2
-K-
Mode 1
[AnkleAngle]
u+b
Theta2
K3
-K-
[AnkleVel]
-K-
b2
Add2
Mode 2
K4
[AnkleAngle]
u+b
Theta3
-K-
[AnkleVel]
-K-
b3
Add3
Mode 3&4
Multiport
Switch
-1
Gain
1
Ankle Torque

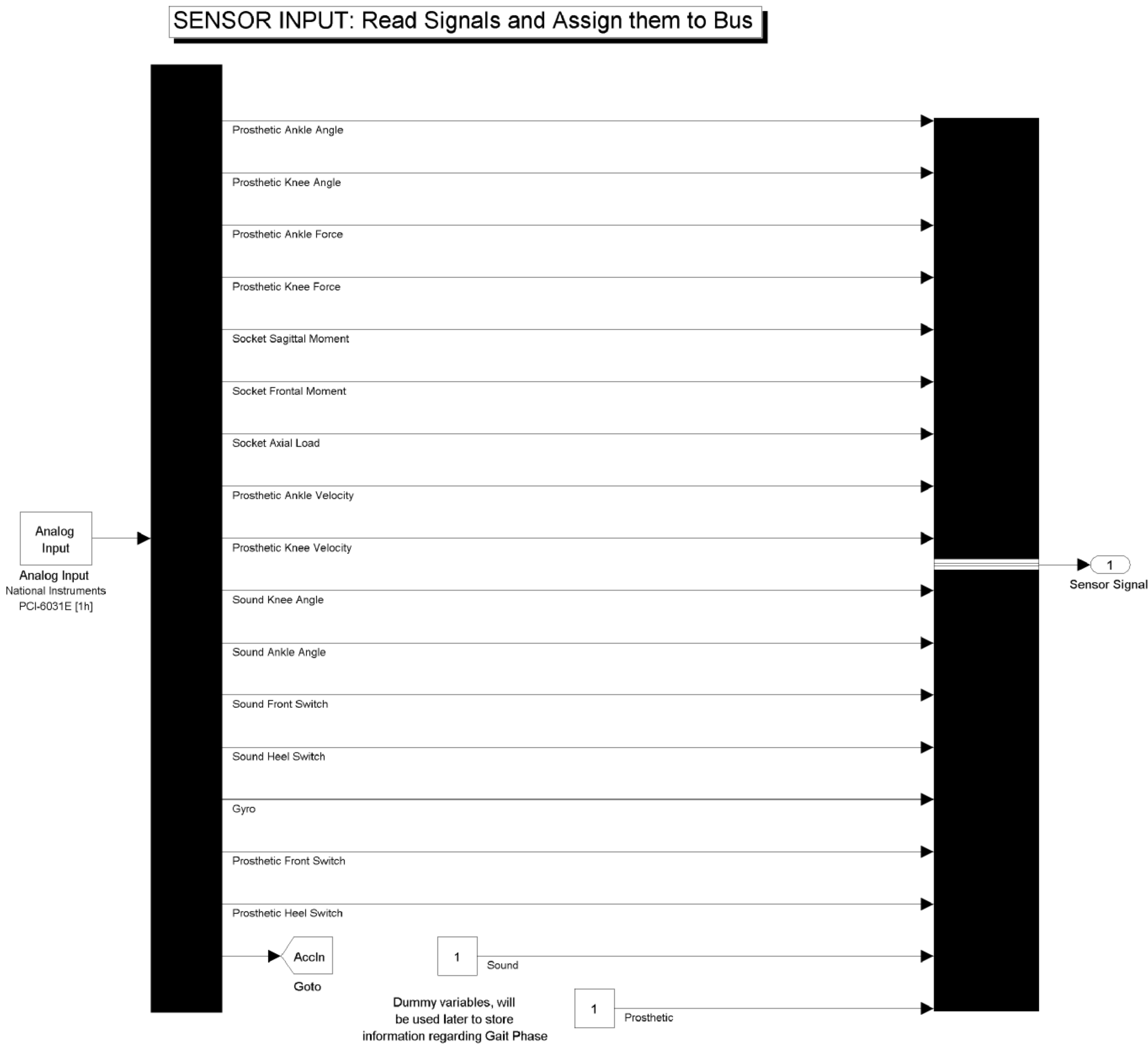
Determine the equilibrium angle:
(1) For Ankle angle < threshold, equilibrium angle = Given Eq. Angle
(2) For Ankle angle > threshold, equilibrium angle = varies according to ankle_angle (positive feedback)




Determine Torque for knee depending upon mode
2
Knee Angle
[KneeAngle]
Goto
3
Knee Vel
[KneeVel]
Goto1
4
Ankle Angle
[AnkleAngle]
Goto2
1
Knee Mode
Mode 0 & 1
[KneeAngle]
u+b
Theta1
K1
-K-
[KneeVel]
b1
-K-
Add
Mode 0 & 1
[KneeAngle]
[AnkleAngle]
Knee Angle
Ankle Angle
Theta
Determine Equillibrium Angle
Add1
K2
-K-
Mode 2
[KneeAngle]
u+b
Theta2
K3
-K-
[KneeVel]
b3
-K-
Add2
Mode 3
[KneeAngle]
u+b
Theta3
K4
-K-
[KneeVel]
b4
-K-
Add3
Mode 4
Multiport Switch
-1
Gain
1
Knee Torque

Determine the equilibrium angle:
(1) For knee angle < threshold, equilibrium angle = threshold
(2) For knee angle > threshold, equilibrium angle = varies according to ankle_angle




SENSOR INPUT: Read Signals and Assign them to Bus
Analog Input
Analog Input
National Instruments
PCI-6031E [1h]
Prosthetic Ankle Angle
Prosthetic Knee Angle
Prosthetic Ankle Force
Prosthetic Knee Force
Socket Sagittal Moment
Socket Frontal Moment
Socket Axial Load
Prosthetic Ankle Velocity
Prosthetic Knee Velocity
Sound Knee Angle
Sound Ankle Angle
Sound Front Switch
Sound Heel Switch
Gyro
Prosthetic Front Switch
Prosthetic Heel Switch
AccIn
Goto
1
Sound
1
Prosthetic
Dummy variables, will
be used later to store
information regarding Gait Phase
1
Sensor Signal

REFERENCES

[1] Winter, D.A., "The biomechanics and motor control of human gait: normal, elderly and pathological," University of Waterloo Press, 2nd ed., 1991

[2] Winter, D. A. and Sienko, S. E., "Biomechanics of below-knee amputee gait," J. Biomechanics. 21, 361–367., 1988.

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