Biologically Inspired Autoadaptive Control of a Knee Prosthesis

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Abstract
This thesis describes the electronic control of a knee prosthesis for amputees. A microprocessor receives data from sensors, processes it and determines the proper level of rotary resistance for the joint. Control is maintained through one of two algorithms -- one for a system with sensors only for knee angle and axial force and one that also senses bending moment at the knee base. The electronic knee can control stance stability, adapt to walking cadence and detect stairs and standing modes, all advantages over the conventional mechanical knee. It will also allow flexion during stance - an important component of normal gait that most prostheses do not allow. Parameters for the knee are set automatically from observing the subject walk, rather than relying on the judgment of the prosthetist doing the fitting. As the subject moves, the output of the microprocessor changes and adapts to the actions of the subject, who might be walking faster, picking up a suitcase, or changing shoes. The algorithms were developed using five amputees with varying physical characteristics. Safety, comfort, and natural-appearing movement were considered in the project.

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Title: Assistant Professor of Electrical Engineering and Computer Science, MIT
Dedicated to Baby Lea, her extended family, and her great-great grandfather
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Chapter 1-Introduction

Prostheses are artificial replacements for body parts lost due to injury or illness. Knee prostheses in particular, are used by subjects who have their leg amputated between the knee and the hip. Common causes include accidents, military losses, and diabetes. In the United States alone there are tens of thousands of people suffering from above the knee (or through the knee) amputations every year (20).

Knee prostheses have been designed for thousands of years. The earliest involved simply a stick to walk on. Later, a hinge was introduced (often two pieces of wood bound by cloth) to allow the knee to bend during swing.

More recently, especially in the aftermath of the second world war, more advanced knee units were designed. Many of these newer knees improved upon the concept of a "hinge knee" by adding hydraulic cylinders which could dampen the rotation of the knee by providing a resistive torque about the joint. In some units, this resistance was adjustable by the prosthetist setting up the unit for individual subjects. Adjustments were made based on the
individual prosthethist's judgment, training, and experience.

In the very recent past, research has been carried out in designing an electronically controlled knee. Electronic knees use some form of computational intelligence to control the resistive torque about the knee. There are several potential advantages to electronic knees over the "conventional" designs. Electronic knees can be programmed to detect stumbles and other pathological behaviors and react appropriately. They can provide a more natural stance phase of gait by discriminating between early and late stance using sensor information. They can be designed so as to give different levels of damping during swing so as to optimize for different walking speeds (assuming they have the appropriate sensors and a method for estimated walking speed).

With the correct control algorithm, stairs, sitting down, and other non-standard gait behavior can be detected and accounted for. Finally, it is possible to program such a knee to allow the knee to flex and extend while bearing a subject's weight (stance flexion). This feature of normal walking is not possible with most conventional prostheses.
Several research groups have been involved in designing prototype knee controllers for use in the laboratory. Among them, Bar(30) designed a microprocessor controlled knee based on observing the reaction of the sound side leg and acting accordingly. Aeyels(2,3) worked with electromyographic voluntary control of a knee prosthesis, as did Myers(12,14), Triolo(13), and Peeraer(8). Aeyels, along with Peeraer, also did preliminary work on including stance flexion in a prosthesis(4,9). Popovic(5,7) worked on using output space Lyapunov tracking for control of a knee prosthesis while Ju(3) attempted to use “fuzzy logic” for the same purpose. Wang(6) did simulations of adaptive control. Chitore(10) and Nakagawa(11) also worked on the control of an electronic knee. Kautz(15) designed a knee based on input from the sound side leg (“echo” control). Our control algorithm followed a more computationally simple strategy than many of these and did not involve either myoelectric sensors or sensors on the contralateral leg.

Flowers, at the Massachusetts Institute of Technology, and his students worked on a variety of microprocessor based knees for use in the lab. Two of
his students, Grimes (19) and Darling (18) both worked on controller designs based on the concept of “echoing” the actions of the sound leg with the prosthesis. Two others, Qi (17) and Goldfarb (16) developed a multi-mode controller using only sensor information from the prosthesis side in the control algorithm. Their progress heavily influenced the initial paradigm of the design presented in this work.

A small number of companies have also developed electronic knees for clinical use. Prominent among these are the Endolite Intelligent Prosthesis and the Otto Bock C-leg (23). The Endolite IP allows the prosthetist to set the resistance for three different speeds of walking for both flexion and extension in swing (see chapter 2). The Otto Bock C-leg also provides adjustable resistance for flexion and extension in swing. The onboard intelligence can detect when the user is descending stairs or sitting down. In addition, it allows the prosthetist to adjust the resistance during the stance phase of gait. A special software package is necessary for the prosthetist to calibrate several parameters for each subject.
The new M.I.T. knee described in this thesis attempts to take full advantage of the possibilities of an electronic knee. It estimates walking speeds and separately and automatically adapts swing resistance for each walking speed. It also adjusts resistance during stance phase based on the subject's weight and walking speed. It can detect stairs and sitting behavior. Finally, it is designed to be self-calibrating. All the parameters for optimal knee resistance and detecting switches in modes of gait are automatically determined by the processor from sensor data collected when the subject walks.
Chapter 2 - Normal and amputee gait

Walking for the able-bodied is easy to do. It is therefore easy to overlook how many complex, synchronized actions are necessary. To further complicate the matter, it is very difficult to define "good" versus "bad" walking behavior in quantitative terms. Consider, for example, someone walking with a rock in his shoe. A typical observer will immediately be able to notice that something "looks wrong" about the way the person is walking. But describing the abnormality quantitatively, in terms of the joint angle trajectories of the hips, knees, and ankles, would generally be very difficult. Likewise, if the joint angles were recorded on a computer, someone looking at them might detect nothing wrong with the gait.

For the design of a knee prosthesis, this problem is compounded by only having data from the knee. No data is available about the motion of the ankle, hip, or center of mass. Likewise, sensor information is only available from the leg with the prosthesis. Since much of "good" walking is a function of symmetry
between the two legs, the lack of the information from the contralateral leg is a serious difficulty.

Many studies have been done of both normal and amputee gait (24-29,31,32). Figure one shows a schematic of a normal gait cycle. Gait is conventionally divided into several phases by both kinematic and kinetic barriers.

A gait cycle starts at heel strike (when the foot hits the ground). From heel strike until the knee finishes flexing, is called "early stance" or "stance flexion". During this phase, the subject's weight is loaded on the leg and the knee flexes. The purpose of stance flexion is to absorb some of the shock of heel strike. Most sources also claim that it reduces the vertical gyrations of the center of mass. Without stance flexion, there would be a large difference in the body center of mass between when the leg is supported over one straight leg and when the body is
supported equally on two legs. Recent research however has questioned whether the timing of stance flexion leads to any actual improvement in the trajectory of the center of mass (1).

From the time of maximum knee flexion until the knee extends back to straight is described as "late stance" or "stance extension". The power to straighten the leg comes from the hip and/or the alternate leg pushing off, and/or the forward momentum of the body.

After the leg straightens during stance it begins to flex again in preparation for the swing phase.
This phase, until the foot leaves the ground, is known as "pre-swing". Pre-swing for one leg begins slightly after heel strike of the opposite leg. The time when both legs are on the ground and supporting the body is referred to as "double support".

From the time when the leg leaves the ground until it reaches its maximum flexion angle is referred to as "swing flexion". Flexing the leg (and therefore shortening it) is important so as to prevent the toe from hitting the ground as the leg swings forward in preparation for the next heel strike. Too much flexion, however, will take time and not allow the leg, when it swings forward, to be extended in time for the next heel strike.

From the time of maximum flexion in swing to the following heel strike is referred to as "swing extension". Ideally, the leg straightens out at the same time the foot is ready to contact the ground.

A gait cycle for an amputee using a "conventional" prosthesis looks much like that shown in figure one. The major phases are the same as for normal gait except that conventional prostheses do not allow stance flexion and extension. Prosthetic gait
also tends to be asymmetric between the prosthetic leg and the biological leg.
Chapter 3 - The Knee

This chapter will provide a brief overview of the hardware on which the controller is run. Only those aspects of the design necessary for explanation of the control will be detailed. An above-the-knee amputee uses a prosthesis system containing three basic components - the socket, the knee, and the foot/ankle unit. Although some systems come with two or more of these components combined, generally they are modular and can be intermatched to meet the needs of the prosthetist and subject.

The firm, comfortable fitting of the socket to the subject's stump is probably the most important factor in successful prosthesis-aided walking. Sockets are designed by taking an impression of the subject's remaining leg stump and then fitting a carbon composite mold. Most sockets stay attached to the stump due to suction caused by a vacuum in the part of the socket not filled by the stump. The knee unit bolts to the bottom of the socket. In addition to the socket fit, the length of the remaining leg stump can have a large effect on how well an amputee can walk. A longer stump generally means more useful
remaining muscle mass and a better lever arm to control the lower leg prosthesis.

Ankle-foot units come in three basic varieties—solid, hinge, and energy storage modules. An example of a solid ankle-foot unit is the SACH (Sold Ankle, Cushion Heel) model. As the name implies, these have a softer ankle built into the rubber foot to allow for some shock absorption at heel strike. The ankle, however, is rigid.

Both hinge ankles and energy storage ankles, on the other hand, allow for flexion and extension of the ankle. Hinge ankles allow free rotation about the ankle joint and are not often used. Energy storage units, on the other hand, have springy elements in the ankle (and/or foot) which are bent and then release their energy in pre-swing. Since in normal walking, much of the energy comes from the ankle, these units have become increasingly popular.

The MIT knee itself (see figure two) is mechanically passive. It generates no energy and can only resist applied torques (this is true of all knees currently available). The basic knee design is a series of interspaced blades which slide past each other when the knee rotates. An electromagnet
provides a varying magnetic field perpendicular to the plain of the blades. This field pushes the blades together and makes it harder to slide past one another. The knee is also filled with a magnetic particle fluid. The field lines up the particles of the fluid into chains and thus changes the shear resistance of the fluid. The result of this design is that the knee will provide a resistive torque as a function of current applied to the electromagnet (see figure three).

Figure 2- The M.I.T. Knee (connected on the top to a socket to hold the leg stump and on the bottom to an artificial leg, ankle, and foot)
Figure 3- Knee resistive torque as a function of commanded current. Several data points were taken at each current to check repeatability.

The knee has a built-in rubber "knee-cap". Its purpose is to dampen somewhat the noise and vibrations when the knee extends rapidly to a completely straight configuration.

There are three input signals to the knee: knee angle, bending moment below the knee (either a loaded toe or loaded heel), and applied axial force to the knee (force along the axis of the leg). The derivative of knee angle is also taken to give angular velocity. The electronic controller of the knee consists of a microprocessor, memory units, analog to
digital converters for the sensor, and a current output to the knee actuator.
Chapter 4- The Controller

Conventional Prostheses

Any conventional knee prosthesis has four major goals corresponding to four of the five phases of gait (there is no requirement for stance-extension since conventional prostheses do not allow the knee to flex and extend during stance). During early stance it must provide stability (i.e. prevent knee buckling). During pre-swing, the knee must go easily into flexion. During swing-flexion, the maximum heel-rise must be limited. Finally, during swing extension, there must be sufficient deceleration for a soft stop while still ensuring the knee reaches full extension.

For conventional prostheses, the first two goals are achieved by altering the static alignment of the subject's weight line relative to the axis of rotation of the knee (see figure four). If the weight line is anterior to the knee, the knee is said to be stable (that is, it will not buckle). If the weight line is posterior to the center of rotation at heel strike, the subject must provide an extensive torque about the knee with their hip muscles to stabilize it. The
subject's ability to stabilize the knee in this fashion determines, when aligning the knee on the subject, how far posterior the weight line can be to the axis of the prosthesis.
More stable alignment

Less stable alignment

Figure Four- Stability in the alignment of a prosthesis. In A, the leg is in the position it would be at heel strike (initial loading). \( M_v \) is the voluntary hyperextensive moment the subject can supply to keep the knee from buckling. The weight line, \( P \) causes a flexive torque about the knee causing it to buckle. The sum of these two torques can be represented by an equivalent force vector \( Q \). \( Q \) must pass in front of the knee axis for the knee to be stable at heel strike. The magnitude of \( M_v \) determines how far back, relative to the knee axis, the socket can be aligned. In C, the leg is in the position it would be in during pre-swing. In this diagram, \( M_v \) is the voluntary flexive torque that can be supplied by the subject (from his hip muscles). The weight line, \( P \), will tend to cause a hyperextensive torque about the knee axis. The equivalent force vector, \( Q \), must be behind the knee axis for flexion to take place leading to swing. The diagram on the left shows how the further forward the socket is aligned relative to the knee axis, the more hyperextensive torque is supplied by the weight line at heel strike. (34)

When attempting to go into pre-swing, there must be little or no resistive torque to interfere. If the weight line of the subject at pre-swing falls in front of the axis of the knee, an extensive torque is generated which makes pre-swing more difficult. Knee alignments with the weight line relatively far forward (sacrificing ease of pre-swing for stability at heel strike) are referred to as "safe" alignments. Knee
alignments with the weight line relatively far back (sacrificing stability at heel strike for ease of pre-swing) are referred to as "triggered" alignments.

To provide flexion and extension resistance during swing, most prostheses have hydraulic cylinders of some sort\(^1\) to aid with swing control (see figure five). The hydraulics provide a resistive torque proportional to the velocity of the knee squared. On some knee units, the damping constant can be set by the prosthetist. Some units also allow the flexion and extension damping to be set separately (using different hydraulic cylinders or a set of hydraulic one-way valves).

\(^1\) There have actually recently been introduced several new varieties of mechanical knees including those with moveable centers of rotation and "friction locks" to enhance stance stability. A detailed listing is beyond the scope of this work, but can be found in (33)
The MIT knee

The knee control algorithm takes information from the sensors and from its internal state. It then determines which phase of gait the walker is in. Based on the phase of gait and the sensor inputs, it determines the appropriate resistance for the knee.

The controller implements a state machine with each state corresponding to a phase of gait (see figure six). State one is stance-flexion (or early-stance), State two is stance-extension (or late stance), State three is pre-swing, State four is swing-flexion, and State-five is swing-extension.
Stance

The MIT knee is programmed to tell the difference between initial loading during stance and the beginning of pre-swing. It can therefore provide stance stability without needing to rely solely on static alignment and subject hyper-extensive effort like conventional prostheses. The knee therefore does not have to be aligned such that it will be locked all the way through stance to ensure safety. Rather, it is designed to allow the controlled flexion and extension during stance present in normal gait.
Figure 6 - Sensor data from the prosthesis for a single stride showing knee angle (in degrees), Force (in arbitrary units) and Moment (in arbitrary units). Heel Strike (when the foot first hits the ground) and Toe-Off (when the foot leaves the ground for swing) are marked, as are the five states the controller cycles through during gait.

The first state in the controller corresponds to early stance. The trigger to enter State one is the foot making contact with the ground, as measured by the force sensor. Under normal operation, the knee will flex while in this state and then switch to State two (stance extension) when maximum stance-flexion is reached.
The resistance for early stance is determined as a linear function of the peak force in the step before, as well as angle and angular velocity. That is:

\[ \text{Torque} = B_{\text{early stance}} \cdot \text{angular velocity} \cdot \text{angle} \]

(Where \( B_{\text{early stance}} = C \cdot \text{peak axial load from previous step} + D \))

The logic behind this is that a heavier subject, or a subject walking more quickly (corresponding to a higher dynamic ground reaction force) will require more support from the knee during stance flexion. The constants in the formula (C and D) were derived empirically from testing what levels of resistance were most comfortable for the subjects in the study.

The velocity dependence in the determination of torque is there primarily so that the torque will be zero when the knee starts flexing to allow the subject to initiate bending and zero when it reaches maximum flexion, thereby allowing the subject to reverse and enter stance extension.

The angle dependence in the determination of torque is also there to ensure low torque to allow the beginning of flexion. It also serves to “ramp up” the
torque as the subject bends, giving the more safe feeling that it will catch them.

The second state corresponds to extension during stance. The state begins when stance flexion is completed and the velocity turns positive\(^2\). It ends when the conditions for pre-swing are met.

The resistance for stance extension is a function of the angular velocity, specifically:

\[
\text{Torque} = B_{\text{late stance}} \times \text{angular velocity}
\]

The damping constant, \(B_{\text{late stance}}\) is fairly low and was determined empirically from testing for comfort on the subjects. The only major issue with this state is preventing noise when the knee re-extends into the "knee-cap", which is already partially limited by the rubber bumper (see chapter 3). Therefore, no adaptation, per se, is necessary for this state. As in State one, the primary reason for the velocity dependence is to assure that at the beginning of the state (when the knee is maximally flexed) the torque will be low, allowing the subject to begin extension.

\(^2\) In all places in this document, 0 degrees refers to the knee when in is straight. By convention, positive angles refer to the knee in flexion and positive angular velocity refers to a flexing velocity.
The third state is pre-swing. The trigger for pre-swing presents a relatively difficult problem. At heel strike, when high torque is needed to prevent knee collapse, the knee is straight and still with the subject's weight on the leg (see figure six). Likewise, at the beginning of pre-swing, the knee is straight and still with the subject's weight still on the leg. In pre-swing, however, it is critical to have no resistive torque to impede the subject's kicking the leg out into swing phase. Therefore one is left with two conditions with very similar sensor signals but requiring diametrically different responses.

Two different controllers were developed for this project, one for a two sensor system- angle and force, and one for a system with three sensors- angle, force, and bending moment (see chapter 3). A different solution to this state transition problem was found for each of these systems.

For the two sensor system, four criteria need to be met for switching from State two to State three.

a) The knee must be close to straight (within 2 degrees)

b) The knee must be still (angular velocity of
c) The axial force must AT SOME POINT have passed higher than a force threshold

d) At least 300 msec must have elapsed since 'c'

Condition 'a' is present so that the knee can not go into low torque mode when the subject has a bent knee and is relying on it for stability. The angle is not set to zero (perfectly straight) since with stance flexion and extension, it is possible to choose to go into pre-swing before stance-knee-extension is entirely complete.

Condition 'b' is present so that if the knee IS going through knee flexion and extension, pre-swing will not be triggered before the knee is fully extended (or as extended as it is going to get) and comes to a stop. Without this requirement, the last two degrees of stance-extension would be in pre-swing at zero torque.

Condition 'c' prevents the knee from going into pre-swing immediately at heel strike. The force threshold is calibrated during the first ten steps of walking by taking the average peak force per step and dividing by 1.2 and then multiplying by 60%.
theory, peak force when walking is approximately 120% of body weight, so the threshold is approximately 60% of body weight. It takes some time from initial heel contact to load to this level. Condition 'd' then gives the subject time to move their weight line forward. By the end of three hundred milliseconds, a subject will have already initiated stance flexion if likely to do so (in which case conditions 'a' and 'b' prevent the transition to pre-swing).

The knee is also designed to provide stability when standing, crouching, or sitting. If State one or State two is entered and the axial force is LESS then 60% of body weight (e.g. they are standing on both feet) pre-swing will not be entered and stance stability will be maintained.

For the three sensor system, the switch to state three is simpler. Three conditions need to be met.

a) The knee must be close to straight (within 2 degrees)

b) The knee must be still (angular velocity of zero)

c) The axial bending moment must indicate a toe-
load above a certain threshold.  

Looking carefully at figure six one observes that stance begins with heel strike (a heel load) and ends with the weight loading the toe. The threshold for condition 'c' is determined during the first ten steps by taking the average peak toe load per step and multiplying by 80%. In the case of standing still, crouching, or sitting, the load should be roughly equally balanced between heel and toe so condition 'c' is not satisfied and stance stability is maintained.

Swing

The swing portion of gait is speed adaptive in two senses. In the immediate sense, resistance from the knee (in most of swing) is proportional to the angular velocity (as is the case with hydraulic knees). Angular velocity dependence is beneficial because a faster moving knee has more kinetic energy and requires more resistive torque to slow it.

At a higher level, the damping parameters for swing are determined separately for each walking

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3 Throughout this work, a toe moment (the moment caused by loading the toe) is given a negative sign and a heel moment (the moment caused by loading the heel) is given a positive sign. Therefore, a large toe moment (as required for the transition from late stance to pre-swing) will be a large negative value.
speed. This is an advantage over mechanical units whose damping level can only be set to one value (and will therefore perform well only over the range of walking velocities for which it was optimized).

There are several possible methods of approximating walking speed. Obviously, the optimal method would be to directly measure the linear speed of the person walking, such as by using an accelerometer and differentiating. Unfortunately the noise inherent in such a system makes it impractical.

Another fairly direct method is to measure the stride time. Shorter stride times will result in faster gait speeds. The major disadvantage of this method is that the information on walking speed is always at least one step old (one must complete a stride before the stride time is known and one can only apply that information to the next step).

One method for approximating the subject’s walking speed before swing phase begins is to look at the peak axial force during stance. It is logical to assume that as walking speed increases, the dynamic component of the vertical reaction force will increase as well (for example, it can reach two or three times the subject’s rest weight when running).
Another possibility for approximating walking speed is to look at the contact time for the stance phase preceding a given swing phase. Biological data (see chapter 6) shows a strong inverse correlation between contact time (time from heel strike to toe off) and walking speed (see figure seven).

![All Normal Subjects: Contact Time](image)

**Figure 7** Contact time (from heel strike to toe off) vs. walking speed for a pool of unimpaired subjects

To test for the best possible discrimination between speeds, a subject was asked to walk fast, medium, and slow. For several steps at each speed, the maximum stance axial force and stance contact time were recorded. The results can be seen in figure eight. Based on this data, there is a clearer
distinction between the three walking speeds using contact time then using axial force.

Another basic premise for swing is that the resistance should be applied as late as possible in the swing. This is based both on mimicking biological behavior and for safety reasons.

The fact that biologically, muscle moment is generally applied only at the beginning and end of swing phase can be seen from figure nine. From a more practical standpoint, both consultation with professional prosthetists (20) and our own experience shows that torque applied too early leads to at least subject discomfort and at worst tripping.

The fourth state is swing flexion. It begins when the axial force sensor determines that the foot is on the ground and ends when the knee reaches its maximum flexion in swing and changes direction.
Figure 8- Two methods of estimating walking speed. Both the axial force of the leg and the contact time are correlated with speed. There is less overlap and therefore more accurate detection using contact time.
Figure 9- Knee moment in biological walking shows that most work in swing is done at the end of swing flexion. From (31)

Knee resistance for swing flexion is a function of angle and velocity. There is no torque at all until the knee flexes past 15 degrees. After that, it is determined by:

\[ \text{Torque} = B_{\text{swing flexion}} \times \text{angular velocity} \times (\text{angle} - 15) \]

The dependence on angular velocity, as previously mentioned, is useful because within a given step, there is built in error correction. A knee flexing
faster than expected for a given walking speed will require more damping to slow it in time.

The angle dependence prevents resistive torque from being applied too early, as previously discussed\(^4\). The form of the function also prevents torque from being applied too suddenly (which the subject could feel as a jarring sensation). Specifically, the maximum possible damping value ramps as a linear function of flexion angle, starting at fifteen degrees.

The auto-adaptation for State 4 is designed to limit the maximum flexion angle for swing. The peak flexion angle (and as a consequence the maximum heel rise) during swing is very important in amputee gait. If it is too high, the prosthesis will take too long to complete the swing cycle and will not be extended and ready for the next heel strike. To prevent tripping in this way, amputees are forced to either walk slower or work harder to push the knee forward during swing extension.

If, on the other hand, the heel does not reach a sufficient angle, there is an increased chance that

\(^4\) For some walkers, the knee flexes past 15 degrees while the leg is still on the ground (i.e. the controller is still in State 3- pre-swing). In this case, torque is still applied in pre-swing as though it is in State 4.
the amputee will stub their toe on the ground when swinging it forward (because flexing the knee in swing has the effect of “shortening” the leg to assure toe clearance with the ground).

Finally, a large difference in maximum flexion between an amputee’s biological leg and the prosthesis is clearly visible and does not lend itself to dynamic cosmesis.

Adaptation for State 4 is designed so as to, if possible, keep the maximum knee flexion between sixty and seventy degrees. It should be noted that this is slightly arbitrary. Different works of literature suggest anywhere from 50 to 70 degrees as being reasonable for unimpaired adult walkers. The upper end of the range was chosen because

a) It is safer to have slightly too much flexion then to risk applying too much resistive torque.

b) As discussed in Chapter 6, different studies using different methods for determining angles are difficult to compare.

The adaptation algorithm, therefore, checks for the maximum flexion angle in each step. If it is
greater than seventy degrees, the constant $B_{\text{stance flexion}}$ is increased by an amount proportional to the amount by which it is greater.

If, after a few steps, the maximum flexion angle in swing is always below sixty degrees, the constant $B_{\text{stance flexion}}$ is decreased. It is important to remember that since the knee can only supply resistive torques, nothing can be done for a subject whose maximum flexion angle is less than 60 degrees even when $B_{\text{stance flexion}}$ is zero.

Since the controller is speed adaptive, a different value for $B_{\text{stance flexion}}$ is stored for each walking speed range (as approximated by contact time). These values are stored in an array and the correct one chosen for any given step.

The fifth state is swing extension. It begins when the knee reaches maximum flexion in swing and begins to extend. It ends at heel strike at the beginning of the next stride.

The primary goals are to dampen swing sufficiently such that terminal impact (the impact when the knee reaches 0 degrees and is mechanically forced to stop) is not too hard (the knee is not traveling too fast), while simultaneously not taking
too much energy out of the swing such that the subject has to use excessive energy to extend the leg in time for heel strike.

Knee resistance is a function of angle and velocity, being turned on for only the last few degrees of extension. The torque is governed by the equations:

\[
\text{Torque} = 0 \text{ for knee angle } > X \\
\text{Torque} = B_{\text{swing-extension}} \times \text{angular velocity for knee angle } < X
\]

Adaptation for swing flexion is based on the concept of using enough deceleration to decrease terminal impact while ensuring that the knee always reaches full extension before heel strike. The angle \( X \) at which the damping turns on starts at zero and increases until there is a step in which swing does not quite reach full extension. At that point, the angle \( X \) stops rising.

Experimentation over the course of this study has shown that it is absolutely critical that the leg always reach full extension. If it doesn’t, even by a small amount, most subjects will be extremely uncomfortable when heel strike occurs at an angle.
Therefore, once a single step has not made it to full extension, X will not increase again unless 20 steps do make it to full extension. It only requires a single step not reaching full extension, however, to cause X to decrease. The result is an adaptation biased towards lower torque for safety.

The algorithm as described so far has a problem. The more powerful walkers in the subject pool would, in a desire to **never** let the knee not reach full extension, put more and more energy in from the hip as X increased. There result was that X would continue increasing far past the point where they could walk without quickly fatiguing. Therefore, as an additional safety precaution, there is an empirically determined maximum value for X. It was determined by measuring the highest value that allowed for comfortable walking in the most "sensitive" member of our subject pool.

$B_{swing\_extension}$ is held constant at a fairly high value. The decision to modulate the angle range where the knee is slowed rather than the level of the torque (which is high when it is on at all) is consistent with the principle of using torque only as late in swing as possible.
As in State 4, the controller is speed adaptive. Therefore, a different value for $X$ is stored for each walking speed range. These values are stored in an array and the correct one chosen for any given step.
Chapter 5 - Extra modes of walking and features

STAIRS

Amputees use several strategies for walking down stairs. The easiest method is to lower oneself down to the next step using the biological leg. The prosthetic leg is then brought down to the second step and the process repeated for the third step (see figure 10). The stairs are therefore descended one step at a time.

Figure 10 - Descending stairs one at a time. The prosthesis (the dashed leg) never has to support weight when bent.

Another method for more aggressive amputees is known as "jack-knifing". First the prosthetic leg is lowered to step 2 using the biological leg. Then, rather than bringing the biological leg down to the second step, it is lowered directly to the third (see figure 11).
Figure 11- Descending stairs “step over step”. The prosthesis (the dashed leg) has to support weight when bent.

This requires the subject to bear his full weight on the prosthesis. Since conventional prostheses provide little to no stance torque, this means the knee collapses under the weight. By using excellent timing, an amputee can manage to have their biological leg in place on the third step in time to catch him. The descent however, in addition to being difficult, is noticeably asymmetric.

The MIT knee allows for stance support during stair descent. The difficulty (similar to that in the State 1 to State 3 transition) is that bending the leg to go down stairs looks very similar (to the controller) to the actions leading to pre-swing in level ground walking. In the case of pre-swing, the
leg is straight and still with the subject’s full weight on it just before it begins to flex for pre-swing. The same is true for flexion to lower the body to the next stair. In the case of pre-swing, as previously mentioned, it is critical to have no torque. In the case of stair descent however, a high level of torque is needed.

The two and three sensor systems handle the state transition problem differently. For the two sensor system, the key is condition ‘d’ in the requirements for transition from stance to pre-swing. Experiments in this study show that almost without exception, walking down stairs is fast enough that the knee begins flexing before 300 msec have elapsed from the time the knee is loaded. Once the knee is bent past two degrees, condition ‘a’ keeps it from transitioning to pre-swing. The leg then will leave the step and go into swing phase.

In the event that the system does transition to pre-swing during stair descent, there is a method (using the two-sensor system) to detect the mistake. When entering pre-swing, the force drops off rapidly as the foot leaves the floor for swing. When, on the other hand, the knee is starting to flex to lower the
body down steps, the force falls off more slowly (see figure twelve). Therefore, if the force does not fall off with a certain slope (measured with respect to angle bent in this case), the state machine goes back into State one.

**Figure 12- Axial Force unloading during knee flexion at the end of stance-Stairs vs. slow and fast walking**

In the three sensor system, a large toe moment is one of the triggers to go to pre-swing. In stair descent, the subjects are instructed to place their foot with the heel on the step and the toe hanging over it. Therefore the toe moment is not present and the controller stays in stance-flexion while flexing.
In case the system accidentally goes into pre-swing, it will go back if a heel moment is detected while the leg is still close to straight (when the leg is bent in pre-swing, it means the foot is being unloaded and the moment is no longer a reliable signal (see figure 6)).

The three-sensor system is more robust in terms of state detection for stair descent. The trade off is the additional cost (and/or) weight of another sensor and the added possibility of malfunction should the sensor break.

"Pathological" state transitions

In addition to the normal sequence through the states, there are several abnormal (or pathological) sequences to cover unusual situations, hesitations, and stumbles. The details are:

1. State 1 (early stance) can transition to State 3 (pre-swing) without going through State 2 (late stance). This is because many subjects do not make use of the stance-flexion/stance-extension capability of the knee (see Chapter 6) and therefore will never enter State 2. The
conditions for transition from State 1 to State 3 are the same as those from State 2 to State 3. See figure 13 for an example of the signals resulting from a subject going through the normal gait cycle (as in figures one and six) but without flexing the leg in stance.

2. State 1 and State 2 can transition directly to State 4 (swing flexion). This is for situations where, for whatever reason (as, for example in stair descent) where the foot leaves the ground without first going through the pre-swing state. The trigger is simply the foot leaving the ground.

3. State 2 can transition back to State 1 if the leg begins to flex again while still on the ground and the conditions for pre-swing are not met.

4. In swing, if the leg reaches full extension before heel strike, it is held in full extension until heel strike takes place. In case the leg is on the ground and has not been detected because of sensor noise, this “holding torque” is programmed to have a velocity dependence the same as if it were in State 1.
5. If the system has been in pre-swing for several seconds, it will transition back to State 1.

6. In the case of a stumble (i.e. the foot hitting the ground in the middle of swing phase), one of two things should happen. If the controller is in State 4, there is a path directly to State 1 if weight is placed on the foot. If the controller is in State 5, the normal transition to State 1 occurs whenever weight is placed on the foot. In either case, there is stance torque to aid in stumble recovery.
Figure 13- Sensor data from the prosthesis for a single stride showing knee angle (in degrees), Force (in arbitrary units) and Moment (in arbitrary units). For this stride the subject did not flex the knee during stance.

Overviews of both the two sensor and three sensor state machine rules can be seen in figure 14.
Non-zero axial force

State 1
Stance Flexion

- Nonzero axial force
- Axial force has been past threshold for some time
- Knee is still and almost straight

State 2
Stance Extension

- Knee extending
- Nonzero axial force
- Knee Break conditions not met

State 3
Knee Break

- Nonzero axial force
- Axial force has been past threshold for some time
- Knee is still and almost straight

State 4
Swing Flexion

- Nonzero axial force
- Axial force has been past threshold for some time
- Knee is still and almost straight

State 5
Swing Extension

- Derivative of force with respect to angle or time not steep enough for walking
- Time too long for preswing

Zero axial force

Nonzero axial force

Knee extending

Figure 14A: A diagram of the finite state machine with states in ovals and transition conditions in rectangles. A controller using only two sensor inputs: knee angle and axial force.
Figure 14B: A diagram of the finite state machine with states in ovals and transition conditions in rectangles. A controller using three sensor inputs: knee angle, axial force, and the bending moment below the knee.

**Autocalibration**

In addition to the adaptation in the individual states, some information about the subject and the sensor calibration is calculated by the knee during walking. The average peak force and average peak moment are calculated by recording the peak moment for the first ten steps of normal walking when the system is powered up.
The angle sensor is also periodically auto-calibrated. Every 10 steps the minimum angle reached is checked. If it is not zero, the offsets on the sensor are adjusted.

Power Saving

In order to increase the life of the battery, the electromagnet is shut down completely when unnecessary. Specifically, if the velocity sensor reports that the knee has not moved for three seconds, the electromagnet is shut off, regardless of what the state machine says it should be doing. Movement immediately reactivates the torque.

Chapter 6- Analysis

It should be noted at the outset that the goal of this project was not primarily to develop a knee which would allow "better" level ground walking, which is very difficult to define (see chapter 2). Rather, it was to develop a robust control system for an electronic knee which was auto-adaptive, speed adaptive, and allowed for stable stance flexion and detection of stairs and other modes.
The knee was tested on five experienced above the knee amputees. Table one shows the distribution of the subjects' heights, genders, and conventional prostheses.

**Table 1: Information about subjects**

<table>
<thead>
<tr>
<th>Subject Code</th>
<th>Gender</th>
<th>Height</th>
<th>Conventional Prosthesis</th>
</tr>
</thead>
<tbody>
<tr>
<td>CPL</td>
<td>M</td>
<td>6'-1&quot;</td>
<td>Endolite ESK</td>
</tr>
<tr>
<td>CLH</td>
<td>F</td>
<td>5'-4&quot;</td>
<td>Otto Bock 3R60</td>
</tr>
<tr>
<td>RWE</td>
<td>M</td>
<td>6'</td>
<td>Otto Bock intelligent prosthesis</td>
</tr>
<tr>
<td>JBR</td>
<td>M</td>
<td>6'-2&quot;</td>
<td>Tae Len</td>
</tr>
<tr>
<td>LAC</td>
<td>F</td>
<td>5'-5&quot;</td>
<td>Endolite ESK</td>
</tr>
</tbody>
</table>

The first test, and the most important one, was almost entirely qualitative. The subjects had to be able to walk both successfully and comfortably. When descending stairs, the knee had to provide sufficient support and when walking up or down ramps the knee had to continue to function normally. With the exception of a few minor problems mentioned in chapter seven, the knee passed these qualitative tests.

Most quantitative testing was done at Spaulding Rehabilitation Hospital in the Gait Laboratory. Data
collection was done with a Vicon analysis system. The system uses reflective markers placed at pre-determined anatomical locations. Based on anatomical assumptions and measurements made on the subject, joint angles and velocities are estimated. Kinematic analysis was done with the use of infrared cameras which record light reflected off the markers. Two force plates on the walkway record forces and moments during the one or two strides when the subject is in range of the camera.

The marker placement system is discussed by Kadaba(24). As is discussed in detail in that work, different placement systems and analysis routines for the kinematic data will lead to different results for measured joint angles. It is therefore, for example, of no use to compare the maximum knee flexion as measured by the Vicon system to the desired sixty to seventy degree range in the knee algorithm. The knee code is adapting so that the peak angle as measured by the potentiometer on the knee will be in the target range. The readings from the reflective marker system at the Gait Lab are not directly comparable (although they are related).
Likewise, the data from the prosthesis cannot be compared literally to published biological norms (unless they were taken using the exact same system). Therefore, the first data analyzed for this study was from 12 healthy, unimpaired adults. Each subject was told to walk at self selected "normal", "fast", and "slow" speeds.

Each of the amputee subjects was then tested twice. For the first test they used their "conventional" prosthesis (as listed in table one). For the second test they walked using the MIT knee. For each of the tests, like the unimpaired subjects, they walked at three self selected speeds- "normal", "fast", and "slow". For each speed the subject took enough steps such that there were around ten good recordings for both the prosthetic leg and the biological leg.

Among the data recorded on each subject was the walking speed, the peak swing flexion angle, and the time for swing flexion and extension. The data was analyzed so that the subjects' conventional prosthesis and the behavior to the MIT knee could be compared both to the behavior of their biological leg (since
symmetry is very important) and against the range of biological behavior for adults without prostheses.

Data on stair descent was also taken using video at 30 frames per second. The quantitative value being judged was symmetry in time between the sound side and the prosthesis.

Prosthesis induced pathologies

In judging a prosthesis it is critical to remember that there is a human in the system. The MIT knee is designed to adapt its parameters to meet the needs of individual subjects. Subjects, however, will also adapt to compensate for the behavior of the prosthesis they are using.

There are certain common habits that long-time amputees tend to pick up to compensate for deficiencies in the prosthesis. According to a study done by James(29), a normal walker will spend 61% of a walking cycle in stance. An amputee, on the other hand, will spend only 57% of a walking cycle in stance when weighting his prosthesis, but 65% when weighting his natural leg. This is presumably to some degree due to discomfort in putting weight on the prosthesis. Likewise, an average amputee at a self selected
walking speed will walk 0.96 meters per second as opposed to an unimpaired walker who will go approximately 1.5 meters per second under the same conditions. Prosthesis swing is also slower than the swing of a natural leg. Whether this is because the leg reaches a higher maximum flexion angle in swing is unclear.

Another habit many amputees pick up is raising themselves up on their good leg while going over it. Generally, this is because of a fear that the prosthesis, when swinging through, will not clear the ground. Likewise, an amputee will often circumduct (swing the leg around outward) with the prosthesis to make ground clearance more likely.

Most complicating for the purposes of study is the way amputees are trained to pull themselves over the leg (apply a hyperextensive torque) at heel strike. Since most prostheses will not support stance flexion, amputees are trained that any bending of the leg during initial loading is an indication that the knee is about to buckle and drop them.

It is therefore very difficult to convince subjects to allow the knee to flex at heel strike, even after demonstrating (using parallel bars) that
the knee will not buckle and drop them. In biological walking, the knee flexes in stance to somewhere on the order of 20 degrees (depending on walking speed). Only two of the five subjects in this study allowed any stance flexion at all, and only one of them allowed stance flexion above a couple of degrees (that subject’s data is shown in figure 6). Aeyels experienced similar difficulties in his studies of stance flexion on a prosthesis (4).
Chapter 7 - Results and Conclusions

Qualitatively, the knee seems quite successful with the exception of the smoothness of stance flexion (see Appendix A). The stance flexion function was very exciting for the one subject who managed to take advantage of it.

The next test was to check the adaptation routines. An example of swing flexion as a subject walks at a single speed is shown in figure fourteen. Note that the damping factor, $B_{swing\_flexion}$ (see Chapter 4), stays at zero for the first several steps while the sensors auto-calibrate themselves. During this time, heel rise is excessively high (the target being between sixty and seventy degrees). Then, as the adaptation begins, it quickly reaches and maintains a value which puts the maximum swing flexion angle in the target range.

---

5 Note that the angle measurements in figure 14 were measured by the knee's angle sensor, not kinematic measurement system at Spaulding rehab hospital. It is the only set of angles so measured in this chapter (see chapter 5 for details on angle measurements).
Figure 14- Adaptation of swing-flexion damping and the resultant maximum knee-angle as a function of steps taken at a given speed (the target of the adaptation is to keep the maximum knee angle between sixty and seventy degrees)

On stairs, the descent appears much more symmetric than the traditional stair-over-stair "jackknife" descent (see chapter 5). For a quantitative analysis, a subject (the best stair walker) practiced walking on both her normal prosthesis and the MIT knee. A videotape (30 frames/sec) was then taken of her descending a flight of stairs in each. For each step, the time from heel-strike to maximum flexion of the knee was calculated. For her conventional prosthesis, this time, on average, was 25% longer for the prosthesis than for
her biological leg. With the MIT knee, the time for the prosthesis was, on average, only 8% longer than for her biological leg.

As previously mentioned, the subjects at the Spaulding gait lab were asked to walk at a normal pace as well as a fast and a slow one. Figures fifteen through seventeen show the results of that data.
Data on maximum knee angle in swing (figure 15)

The first data observed from the Spaulding gait lab was maximum heel rise in swing. For each subject there are two plots. The first shows the data from the subject walking with the MIT prosthesis. Data from both the prosthesis and sound leg are plotted. The second shows the data from the subject walking with their "conventional" prosthesis. Again, data from both the prosthesis and sound leg are plotted. On both plots, the range of biological data from unimpaired subjects is plotted.

Of the five subjects, two (CLH and JBR) had flexion angles either at or below biological norms with both prostheses. Two of the subjects (CPL and LAC) had unusually high maximum knee angles at fast walking speeds with their conventional prosthesis that were closer to the biological norm with the M.I.T. knee. One subject (RWE) had unusually high maximum knee angles at fast walking speeds both with his conventional prosthesis and with the MIT knee. Presumably he was putting in enough power through the knee that even with maximum resistive torque he was able to overpower it.
Figure 15 - CPL - Swing maximum angle for the MIT and conventional prostheses, the subject’s sound side leg, and reference data from the database of unimpaired walkers
Figure 15 - CLH - Swing maximum angle for the MIT and conventional prostheses, the subject's sound side leg, and reference data from the database of unimpaired walkers
Figure 15 - RWE- Swing maximum angle for the MIT and conventional prostheses, the subject's sound side leg, and reference data from the database of unimpaired walkers.
Figure 15: Swing maximum angle for the MIT and conventional prostheses, the subject’s sound side leg, and reference data from the database of unimpaired walkers.
Figure 15 - LAC - Swing maximum angle for the MIT and conventional prostheses, the subject’s sound side leg, and reference data from the database of unimpaired walkers
Data on swing flexion time (figure 16)

Swing flexion is defined here as the time between when the foot leaves the ground and when it reaches its peak angle in swing. As has been previously noted, prosthesis swing times tend to be slower than swing times from unimpaired subjects. The situation is worse when the maximum knee flexion is too high with the prosthesis. Both subjects (CPL and LAC) who had unusually high maximum swing flexion angles with their conventional prosthesis and improved maximum flexion angles with the M.I.T. knee also showed an improvement (faster and closer to biological times) in the time for swing flexion at fast speeds (where the difference in maximum knee flexion was greatest).
Figure 16- CPL- Swing flexion time for the MIT and conventional prostheses, the subject's sound side leg, and reference data from the database of unimpaired walkers
Figure 16 - CLH- Swing flexion time for the MIT and conventional prostheses, the subject’s sound side leg, and reference data from the database of unimpaired walkers
Figure 16 - Swing flexion time for the MIT and conventional prostheses, the subject’s sound side leg, and reference data from the database of unimpaired walkers.
Figure 16 - JBR- Swing flexion time for the MIT and conventional prostheses, the subject's sound side leg, and reference data from the database of unimpaired walkers
Figure 16- LAC- Swing flexion time for the MIT and conventional prostheses, the subject’s sound side leg, and reference data from the database of unimpaired walkers
Data on swing extension time (figure 17)

Swing extension is defined here as the time between when the leg reaches its maximum angle in swing and when it reaches full extension before heel-strike. Like swing-flexion, swing extension tends to be slower with prostheses than with unimpaired walkers. There did not seem to be a significant difference in swing extension times between the MIT knee and the subjects' conventional prostheses.
Figure 17 - CPL - Swing extension time for the MIT and conventional prostheses, the subject’s sound side leg, and reference data from the database of unimpaired walkers.
Figure 17 – CLH- Swing extension time for the MIT and conventional prostheses, the subject’s sound side leg, and reference data from the database of unimpaired walkers.
Figure 17 - RWE- Swing extension time for the MIT and conventional prostheses, the subject's sound side leg, and reference data from the database of unimpaired walkers.
Figure 17- JBR- Swing extension time for the MIT and conventional prostheses, the subject's sound side leg, and reference data from the database of unimpaired walkers.
Figure 17- LAC - Swing extension time for the MIT and conventional prostheses, the subject’s sound side leg, and reference data from the database of unimpaired walkers
Conclusions

The MIT knee seemed to perform at least as well as the subjects' conventional prostheses in level walking. In addition, its speed adaptive routine seemed to limit excessive swing flexion at fast walking speeds in the majority of trials. Symmetry with the natural leg was not significantly improved, giving much credence to the possibility of using sensors on the other leg for some form of modified echo control in future versions. The stance flexion torque was also not smooth (see Appendix A).

As a multi-mode controller, the knee seemed capable of detecting stairs and when subjects were sitting down. Stance flexion adaptation, unfortunately, could not be fully developed because of subject reluctance to allow it.
Future Work

When a subject bent the knee when loaded, there was a "chattering" effect: That is, the knee torque turned on and off at somewhere between 10 and 30 Hz. In these modes, the knee torque was a closed loop function of velocity (see chapter 4) and thus was effectively acting as a mechanical dashpot (Torque= B* angular velocity). As is well known from control theory, such a passive device cannot be unstable for any value of B. However, the fact that the actuator does not have a perfect response and the delays inherent in a digital system make instability a possibility(35). In effect, these delays limit the gain-bandwidth product of the damping. A compromise was reached by filtering the output torque with a low pass filter. Naturally, however, this adds delay to the response and is thus not entirely satisfactory. More work needs to be done for the next version to identify and model more precisely the source of the instability.

The second problem with stance instability was an inability to come up with a good, autoadaptive controller based on biological walking principles.
Unfortunately, with the subjects’ years of training to not allow knee buckling, it wasn’t possible to get enough data to do so. Assuming that given time with the knee subjects will learn to allow stance flexion, there is an excellent candidate for a very simple adaptation for stance flexion that might be exploited. Namely (as seen in figure 18), in normal walking, the time for stance flexion changes very little with speed (relative to the total time the foot is on the ground, which changes drastically).

![Graph showing walking speed variation of total stance time and stance flexion time for unimpaired subjects.](image)

*Figure 18*
A simple adaptive rule would be to increase or decrease the stance-flexion damping until the time for stance flexion reaches approximately 250 msec.
Appendix A- control code

The function knee_controller1() is called after the sensors are read and before the data is current is output to the knee. There are two hardware gains on the angle sensor which are combined into one variable (ls.qknee_joint) before knee_controller1() is run. There are also two velocity hardware gains. The variable ls.qdknee_joint_hi (the high gain signal) is used for everything except velocity dependence in swing (which uses ls.qdknee_joint_low so the signal doesn’t clip). The output is in the variable ls.torq_out, which ranges from 0 to 120. If the flag ls.enable is not set to one, however, there is no output independent of ls.torq_out. Other variables of note are ls.t (the time), ls.bin (the walking speed range- there are 20 of them calculated based on the contact time to 15 degrees of preswing), ls.force_sens (the axial force), ls.moment (the moment) and ls.state_var (the variable controlling which state the system is in).

/*File: test_knee.c*/

***************************************************************************
* *
#define PI 3.14159265
#endif _6812PLATFORM

/* use 6812 #includes */

#include "usercode.h"
#include <float.h>
#include <math.h>

#else

#include <config.h>
#include <walker/device.h>
#include <walker/dram.h>

#define ANALOG32_LOCAL
#include <walker/analog32/analog32.h>

#include <c3x/c32.h>
#include <math.h>

#include "dsplegui.h"
#include "creature.h"
#include "control.h"

extern DEVICE *dev;

#include "test_knee.h"
#endif
#ifdef _6812PLATFORM

/* some system specific stuff */
#define knee_controller1 control_code

#else
#define knee_controller1 test_knee
#endif

#define CONTACT_1 1
#define CONTACT_2 2
#define KB 3
#define SWING_1 4
#define SWING_2 5

#define bound(x, low, hi) ((x) < (low) ? (low) : ((x) > (hi) ? (hi) : (x)))
#define inrange(x, val) (((x) <= val) && ((x) >= (-val)))
#define mymax(x, y) ((x) > (y) ? (x) : (y))
#define mymin(x, y) ((x) < (y) ? (x) : (y))
#define myabs(x) ((x) >= 0 ? (x) : (-x))

#if defined NEED_CHECK
#define zero_epsilon(x) ((x) < -FLT_EPSILON ? (x) : (FLT_EPSILON ? (x) : 0.))
#else
#define zero_epsilon(x) (x)
#endif

extern float *count_arr[22], *sw2ba_arr[22], *sw1bl_arr[22], *steps_arr[22], *sw2g_arr[22], *sw2sf_arr[22], *sw1ap_arr[22], calib_moment_arr[15];

void torqfun(float a_weight)
{
    /* calculate the cl_torque based on the maximum forcesens from the previous step*/
    if((int)ls.step_count>0.5)
    
    ls.cl_gain=ls.calib_weight/ls.cl_torq_slope+ls.cl_torq_intercept;

    int change_gains()
    /* This function controls adaptation and autocalibration. It is run at the transition from swing_2 to contact_1*/
    {
        int loop;
        float temp=0;
        
        if((int)ls.step_count<0.5)
{  
ls.calib_qkjl_min=200.;  
ls.calib_qkjh_min=200.;  
ls.calib_hs_angle_min=200.;  
}

if(ls.angle_peak<=ls.angle_peak_min_cntrl) /*not enough swing to be a step*/  
    return(0);
if(ls.isstep<0.1) /*total swing too long or too short or swl>C1*/  
    return(0);
if(ls.c2_easy_release_f>0.9) /*sitting down*/  
    return(0);
ls.step_count+=1.0;
/*for first 9 steps, calibrate the toe-moment threshold for pre-swing*/
if((ls.step_count < 9.5))  
    {  
        calib_moment_arr[(int)ls.step_count]=ls.moment_min;  
    }
if((ls.step_count>8.5)&&(ls.step_count<9.5))  
    {  
        for (loop=1;loop<=9;loop++)  
            temp+=((calib_moment_arr[loop]));  
        ls.kb_moment=temp/9.0;  
    }
/*dynamic weight for calculating contact 1 torque*/
ls.calib_weight=ls.force_max;
/*every 10 steps, rezero the angle sensor*/
if(((int)ls.step_count % 10)==0)  
    {  
        ls.qknee_joint_off+=ls.calib_qkjl_min;  
        ls.qknee_joint_hi_off+=ls.calib_qkjh_min;  
        ls.calib_qkjl_min=200.;  
        ls.calib_qkjh_min=200.;  
    }
/*after 10 steps start swing calibration*/
if(ls.step_count>10.5)  
    {  
        /*an array of the number of steps in each speed bin*/  
        *(count_arr[(int)ls.bin])+=1.0;
        /*swlap_arr[bin] is the largest angle in swing_1 since it was last reset.  swibl_arr[bin] is the damping value for a given bin*/  
        if(ls.angle_peak>(*swlap_arr[(int)ls.bin]))  
            *(swlap_arr[(int)ls.bin])=ls.angle_peak;
if(ls.anglepeak>70.0)
    *(sw1bl_arr[(int)ls.bin])+=(ls.swl_delu*(ls.angle_peak-
    70.0));
if (((int) (*count_arr[(int)ls.bin]) % 5) ==0)
{
    if ((swlap_arr[(int)ls.bin]<=60.0)
        *(sw1bl_arr[(int)ls.bin])=(ls.swl_deld*(60.-
        (*swlap_arr[(int)ls.bin])));
    *(sw1bl_arr[(int)ls.bin])=mymax(*(sw1bl_arr[(int)ls.bin]),0.);
    *(swlap_arr[(int)ls.bin])=-200.0;
}
/*swing2ba_arr stores the starting angles for swing2 damping*/
/*the numbers start negative then switch to positive when a
single step doesn’t reach full extension. The absolute value is
taken in any event*/
if((ls.cihs_angle<ls.sw2_caught_ang_min))
    {if(((sw2ba_arr[(int)ls.bin])<0.)
     {*(sw2ba_arr[(int)ls.bin])=ls.sw2_delu;
     /***(sw2ba_arr[(int)ls.bin])=mymax(*(sw2ba_arr[(int)ls.bin]),-
     1.0*(20.0-1s.bin)*1.66);}*/;
     *(sw2ba_arr[(int)ls.bin])=mymax(*(sw2ba_arr[(int)ls.bin]),-
     10.0);
    }else
    *(sw2gs_arr[(int)ls.bin])+=1.0;
}else
    {if((*(sw2ba_arr[(int)ls.bin])<0.)
     *(sw2ba_arr[(int)ls.bin])=-1.0*(sw2ba_arr[(int)ls.bin]);
     *(sw2ba_arr[(int)ls.bin])=ls.sw2_deld;
     *(sw2ba_arr[(int)ls.bin])=mymax(*(sw2ba_arr[(int)ls.bin]),0.);
    }/*sw2ga_arr holds the number of good steps for a speed bin that
has already not reached full extension once. There must be 20
consecutive steps of this sort to increase the damping angle on a
bin like this*/
if (((int) (*count_arr[(int)ls.bin]) % 20) ==0)
{
    if(*sw2gs_arr[(int)ls.bin])>=19.5
    {*(sw2ba_arr[(int)ls.bin])=ls.sw2_delu;
/***(sw2ba_arr[(int)ls.bin])=mymax(*(sw2ba_arr[(int)ls.bin]),1.0*(20.0-1s.bin)*1.66);*/
    *(sw2ba_arr[(int)ls.bin])=mymin(*(sw2ba_arr[(int)ls.bin]),10.0);
    *(sw2gs_arr[(int)ls.bin])=0.0;
}
float mypow(float base, int pow)
{
    int loop;
    float temp;

    temp=1.0;

    for(loop=0;loop<pow;loop++)
        temp=temp*base;

    return temp;
}

int setbin()
{
    int loop;

    /*calculate the value for ls.bin based on the contact time*/
    /*lower values for ls.bin mean faster walking*/
    for (loop=1;loop<=19;loop++)
        if(ls.totstance_time>(ls.minstance+(ls.maxstance-
                         ls.minstance)*(float) (loop)/20.0))
            ls.bin=(float) (loop+1);

    if(ls.totstance_time<(ls.minstance+(ls.maxstance-
                        ls.minstance)*0.05))
        ls.bin=1.0;

    if(ls.speed adaptingf==1.0)
        { 
        ls.sw2_bumper_ang=myabs(*(sw2ba_arr[(int)ls.bin]));
        ls.swl_block=(*(sw1bl_arr[(int)ls.bin]));
        }
    
    /*
    Deserialize
    *******************************************************
    **
    CHECK_TRIGGERS() Check the triggers.
    Ex:
    */
case STATE_NUMBER
    check a particular trigger
    if triggered {
        increment the state_var
        exit_routine
    }

******************************************************************************
** */
void check_triggers()
{

    switch((int) ls.state_var){

    case CONTACT_1: /* */

        if(ls.qknee_joint>=ls.c2_easy_release_ang)
            ls.c2_easy_release_f=1.0;

        if (ls.qdknee_joint_hi<ls.clextend_hys)
        {
            ls.state_var = CONTACT_2;
            ls.stance2_entered_f=1.;
            ls.mid=0.0;
        }

        if(ls.moment>ls.my_moment_max)
            ls.my_moment_max=ls.moment;

        if(ls.moment<ls.my_moment_min)
            ls.my_moment_min=ls.moment_min;


        ls.tot_stance_time=ls.t-ls.stance_start;
        ls.stance_to_to_time=ls.t-ls.stance_start;

        if((ls.stair_f==0.)
            &&(inrange(ls.qdknee_joint_hi,ls.cl_kbvel)) &&
            (ls.qknee_joint<ls.cl_kb_angle) &&
            (ls.moment<=ls.kbmoment*ls.clkscl))
        {
            ls.state_var=KB;
            ls.stance3_entered_f=1.;
            ls.start_kb=ls.t;
            ls.mid=0.0;
        }

        if (((int)ls.on_ground) &&(ls.tot_stance_time>0.4))
        {
            ls.state_var = SWING_1;
            setbin();
            ls.mid=0.0;
            ls.start_swing1 = ls.t;
            ls.time_swing_tot=0.0;
            ls.time_extend_tot=0.0;
            ls.angle_peak=0.0;
            ls.time_peak=0.0;
            ls.isstep=1.0;
ls.stair_auto_f=1.0;
ls.sw2_stop_f=0.0;
ls.sw2_stop_hs_angle=-10.0;
ls.time_ps_ex=0.0;
}

break;

case CONTACT_2: /* */

if(ls.qknee_joint>=ls.c2_easy_release_ang)
   ls.c2_easy_release_f=1.0;

if (ls.qdknee_joint_hi > ls.c2_flex_hys)
   {  
      ls.state_var = CONTACT_1;
      ls.mid=0.0;
   }

if(ls.moment>ls.my_moment_max)
   ls.my_moment_max=ls.moment;
if(ls.moment<ls.my_moment_min)
   ls.my_moment_min=ls.moment_min;

ls.tot_stance_time=ls.t-ls.stance_start;
ls.stance_to_to_time=ls.t-ls.stance_start;

if({ls.stair_f==0.0)
   &&(inrange(ls.qknee_joint_hi,ls.c1_kb_vel)) &&
   (ls.qknee_joint<=ls.c1_kb_angle) &&
   (ls.moment<=ls.kb_moment*ls.c1_kb_scl))
   {  
      ls.state_var=KB;
      ls.stance3_entered_f=1.0;
      ls.start_kb=ls.t;
      ls.mid=0.0;
   }

   /*make a time delay to keep the state from chattering back and forth*/
   if (((int)ls.on_ground)) &&(ls.tot_stance_time>0.4))
   {  
      ls.state_var = SWING_1;
      setbin();
      ls.mid=0.0;
      ls.start_swing1 = ls.t;
      ls.time_swing_tot=0.0;
      ls.time_extend_tot=0.0;
      ls.angle_peak=0.0;
      ls.time_peak=0.0;
      ls.isstep=1.0;
      ls.stair_auto_f=1.0;
      ls.sw2_stop_f=0.0;
      ls.sw2_stop_hs_angle=-10.0;
ls.time_ps_ex=0.;
}
break;

case KB:
if(ls.qknee_joint<15.0)
   ls.tot_stance_time=ls.t-ls.stance_start;
else
   setbin();
ls.stance_to_to_time=ls.t-ls.stance_start;
if((ls.ps_start<-10.)&&(ls.qknee_joint>1.0))
   ls.ps_start=ls.t;
if(ls.moment>ls.my_moment_max)
   ls.my_moment_max=ls.moment;
if(ls.moment<ls.my_moment_min)
   ls.my_moment_min=ls.moment;
if (((ls.t-ls.start_kb)>ls.kb_timeout)
         &&((ls.qknee_joint<ls.kb_c1_angle) &&
         (ls.moment>ls.kb_c1_scl*ls.kb_moment)))
   ls.state_var=CONTACT_1;
   ls.mid=0.0;
}
if (((!int)ls.on_ground))
{
   if(ls.qknee_joint<15.0)
      setbin();
   ls.state_var = SWING_1;
   ls.mid=0.0;
   ls.start_swing1 = ls.t;
   ls.time_swing_tot=0.0;
   ls.time_extend_tot=0.0;
   ls.angle_peak=0.0;
   ls.time_peak=0.0;
   ls.isstep=1.0;
   ls.time_peak_f=0.0;
   ls.sw2_stop_f=0.0;
   ls.sw2_stop_hs_angle=-10.;
   ls.time_ps_ex=0.;
}
break;
case SWING_1: /* */
if (((ls.qdknee_joint_hi < ls.sw1_extends_hys) &&
     ((!int)ls.on_ground))
   ls.state_var = SWING_2;
   ls.mid=0.0;
ls.controlled=0.0;
ls.start_swing2 = ls.t;
ls.time_peak=ls.t-1s.start_swing1;
ls.angle_peak=ls.qknee_joint;
}

if (((int)ls.on_ground))
{
  ls.state_var = CONTACT_1;
  ls.mid=0.0;
  ls.stance_start=ls.t;
  ls.isstep=0.0;
  if(ls.sw2_stop_hs_angle> -9.0)
    ls.c1_hs_angle=ls.sw2_stop_hs_angle;
  else
    ls.c1_hs_angle=ls.qknee_joint;
    ls.controlled=0.0;
    ls.stance2_entered_f=0.0;
    ls.stance3_entered_f=0.0;
    ls.stair_auto_f=0.0;
    ls.force_max=0.0;
    ls.moment_min=0.0;
    ls.my_moment_max=0.0;
    ls.my_moment_min=0.0;
    ls.c2_ease_release_f=0.0;
    ls.ps_start=-20.0;
}

break;

case SWING_2: /* */

/*check the slope of the angular velocity to see if the knee has
stopped extending but hasn't hit the ground yet. If so, use THIS
angle when adapting the swing_2 gains*/

if(ls.qknee_joint<=ls.sw2_stop_min_angle)
  if(ls.sw2_stop_hs_angle<(-9.0))
    if(ls.sw2_stop_f==0)
      {ls.sw2_stop_time=ls.t;
       ls.sw2_stop_angle=ls.qknee_joint;
       ls.sw2_stop_f=1.0;
      }
    else
      if((ls.t-ls.sw2_stop_time)>=ls.sw2_stop_interv_time)
        if((ls.sw2_stop_angle<ls.qknee_joint)<ls.sw2_stop_interv_angle)
          {ls.sw2_stop_hs_angle=ls.qknee_joint;
           ls.time_extend_tot=ls.t-ls.start_swing2;
            ls.time_ps_ex=ls.t-ls.ps_start;
          }
        else
          {
          
        

94
if(((int)ls.on_ground))
{
    ls.time_swing_tot=ls.t-1s.start_swing1;
    if((ls.time_swing_tot>ls.time_swing_long) ||
    (ls.time_swing_tot<ls.time_swing_short))
        is.isstep=0.0;
    if(ls.sw2_stop_hs_angle>-9.)
        ls.c1_hs_angle=ls.sw2_stop_hs_angle;
    else
    {
        ls.c1_hs_angle=ls.qknee_joint;
        ls.time_extend_tot=ls.t-1s.start_swing2;
        ls.time_ps_ex=ls.t-1s.ps_start;
    }

    if((1s.adapt_flag>0.9) &&(Us.stair_auto_f==0.) &&
    (ls.stair_f==0.))
        change_gains();

    ls.stair_auto_f=0.;
    ls.state_var = CONTACT_1;
    ls.mid=0.0;
    ls.stance_start=ls.t;
    ls.c2_easy_release_f=0.0;
    ls.stance2Entered_f=0.;
    ls.stance3Entered_f=0.;
    ls.force_max=0.;
    ls.moment_min=0.;
    ls.my_moment_max=0.;
    ls.my_moment_min=0.;
    ls.ps_start=-20.;
}

break;

default:

    ls.state_var=CONTACT_1;

    break;
}
*/

**************************************************************************
**
STATE_MACHINE() state machine.
void state_machine()
{
    switch((int) ls.state_var)
    {
    case CONTACT_1:
        ls.enable=1.0;
torqfun(ls.calib_weight);
if(ls.qknee_joint<=15.)
        {
        ls.gain=ls.c1_gain*ls.qknee_joint/90.;
        ls.torq=0.;
        }
else
        {
        ls.torq=100.;
        ls.gain=0.;
        }
    break;

    case CONTACT_2:
    if(ls.c2_easy_release_f==1.)
    {
    ls.enable=0.;
    ls.torq=0.;
    ls.gain=0.;
    }
else
    {
    ls.enable=1.0;
    ls.gain=ls.c2_gain;
    ls.torq=0.;
    }
    break;

    case KB:
    if((ls.qknee_joint<15.0) || (ls.qdknee_joint_low<0.))
    {
    ls.enable=0.0;
    ls.torq=0.0;
    ls.gain=0.0;
    }
else
    {
    ls.enable=1.0;
    ls.gain=0.;
    ls.torq=ls.swl_block*ls.qdknee_joint_low*(ls.qknee_joint-
    15.0)/55.0;
break;

case SWING_1:
    ls.enable=0.0;
    ls.torq=0.0;
    ls.gain=0.0;

    if((ls.stair_f==0.)
&& (ls.stair_auto_f==0.) && (ls.qknee_joint>15.0))
    {
        ls.enable=1.0;
/*use ls.torq instead of ls.gain to bypass output filter*/
        ls.gain=0.0;
        ls.torq=ls.sw1_block*ls.qdknee_joint_low*(ls.qknee_joint-15.0)/55.0;
    }
break;

case SWING_2:

    if((ls.qknee_joint<ls.sw2_bumper_ang) &&
    (ls.qdknee_joint_hi<=ls.sw2_bumper_vel) && (ls.controlled!=1.0))
    {
        if((ls.stair_f==0.) && (ls.stair_auto_f==0.))
        {
            /*use ls.torq instead of ls.gain to bypass output filter*/
            ls.torq=ls.sw2_block*myabs(ls.qdknee_joint_low);
            ls.gain=0.0;
            ls.enable=1.0;
        }
        ls.controlled=2.0;
    }
else
    if(((ls.qdknee_joint_hi>ls.sw2_bumper_vel) && (ls.qknee_joint<ls.sw2_end_ang)) || (ls.controlled==1.0))
/*hold the knee at full extension so it doesn't bounce back. Add
in stance torque as well in case the heel-strike trigger is
late*/
    {
        ls.enable=1.0;
        ls.torq=ls.sw2_lock;
        ls.gain=ls.sw2_lockv+ls.cl_gain*ls.qknee_joint/90.;
        ls.controlled=1.0;
    }
else
{ 
ls.enable=0.0; 
ls.gain=0.0; 
ls.torq=0.0; 
ls.controlled=0.; 
}
break;
default:
ls.state_var=CONTACT_1; 
break;
}

void outchan1()
{
/*     Calculate desired torque */
/*ls.torq is a constant. ls.gain is multiplied by angular velocity and then filtered to reduce output "chatter"*/

ls.torq_des =
myabs(mypow(ls.qdknee_joint_act_hi,((int)(ls.pow)))*ls.gain;

if(ls.filt_act==1.0)
{
   ls.oldmid=ls.mid;
   ls.mid=ls.control_dt/ls.tp*ls.torq_des+(1.-
ls.control_dt/ls.tp)*ls.oldmid;
   ls.torq_filt=(1.+ls.tz/ls.control_dt)*ls.mid-
ls.tz/ls.control_dt*ls.oldmid;
   ls.torq_out=ls.torq_filt+ls.torq;
   ls.torq_out=bound(ls.torq_out,0.,35.);
   ls.torq_out=zero_epsilon(ls.torq_out);
}
else
{
   ls.torq_out=ls.torq_des+ls.torq;
   ls.torq_out=bound(ls.torq_out,0.,35.);
   ls.torq_out=zero_epsilon(ls.torq_out);
}
}

void knee_controller1()
{
/*do_safety();*/

98
```c
#include _6812PLATFORM
/* PORTP |= 0x10; */
#endif

if(ls.qknee_joint_low<ls.calib_qkjl_min)
    ls.calib_qkjl_min=ls.qknee_joint_low;
if(ls.qknee_joint_hi<ls.calib_qkjh_min)
    ls.calib_qkjh_min=ls.qknee_joint_hi;

if(myabs(ls.qdknee_joint_hi)<=ls.to_vel)
    ls.to_count+=ls.control_dt;
else
    ls.to_count=0.;

check_triggers();
state_machine();

if(ls.torq_filt>ls.torq_filt_max)
    ls.torq_filt_max=ls.torq_filt;
if(ls.mid>ls.mid_max)
    ls.mid_max=ls.mid;

/*check the power timeout conditions*/
if((ls.to_count>=ls.to_time))
{    
    ls.enable=0.;
    ls.toeveron=1.0;
    ls.to_on=1.0;
}
else
    ls.to_on=-1.0;

outchan1();

if(ls.force_sens>ls.force_max)
    ls.force_max=ls.force_sens;
if(ls.moment<ls.moment_min)
    ls.moment_min=ls.moment;

#ifndef _6812PLATFORM
/* PORTP &= ~0x10; */
#endif
```
Title of Study: Mechanism and Control of a Self-Programming Gait-Adaptive Knee Prosthesis
Principal Investigator: Gill Pratt, Ph.D.
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Informed Consent Document

Title of study: Mechanism and Control of a Self-Programming Gait-Adaptive Knee Prosthesis

Principal Investigator: Professor Gill Pratt

Purpose of Study

The purpose of this study is to develop a new prosthetic knee for trans-femoral (above-knee) amputees based on an imbedded microprocessor controller. Our proposed knee may offer the amputee better swing and stance leg control for steady and unsteady walking behavior.

Experimental Protocol

The experiment will consist of three phases for each attempted controller. To start, you will be fitted with a prosthetic socket by a professional prosthetist. You will then be asked to walk with the knee and comment on its performance. Parameters on the controller will be altered to maximize your comfort and efficiency.

You will be given horizontal bars on each side (or will be escorted by a lab assistant to hold your arm) to catch yourself in the unlikely event of falling. The knee will also be equipped with a mechanical failsafe system that will prevent excessive knee flexion in the event the main system fails. A cable from
the knee to a monitoring computer will record data from the prosthetic leg's built-in sensors. A string attached to your back will be played out as you walk to calculate your position. A lab assistant will walk along with you to insure you do not trip on the cables. The electronics in the knee will be low power and will be heavily insulated from you by the prosthetic socket. Your speed will be measured from sensors on the wall as you walk.

Once the best parameters have been obtained, the second phase of the experiment will consist of walking at a steady pace between the parallel bars. For the third phase of the experiment, you will be asked to walk at varying speeds and up and down steps and ramps. The same safety measures and recording protocols will be used for the second and third phases of the experiment as for the first.

We plan to have you in for a series of sessions, each approximately 3 hours in length, approximately once a week for the duration of the experiment. Total time commitment will therefore be on the order of 120 hours.

Throughout the trials, you may be asked questions as to any difficulties you are experiencing walking at different speeds.

If you are female, you will be asked whether you have reason to believe you are pregnant to prevent possible damage from falling to an unborn fetus.

**Risks and Benefits**

As with any prosthetic walking, there is a small risk of falling. We plan to minimize this risk by:

a) Having a mechanical failsafe system on the knee to stop further knee flexion should the knee fail for any reason.

b) Giving you parallel bars (or a nearby lab assistant) to support you in case you feel unstable for any reason.

c) Programming the knee to ‘fail intelligently’ (i.e. either swing freely or stiffen up) if it runs into a situation it doesn’t understand to allow you to right yourself.

The knee itself is entirely passive (i.e. generates no energy). The electronics are low powered and insulated from you by the prosthetic socket. The data collection wires will be kept away from you by a lab assistant to make sure you don’t trip on them.

As with any new prosthetic, there is some risk of skin irritation. If you feel any discomfort, you should immediately tell the investigator at which point experimentation will stop and the professional prosthetist will fix the problem before the next session. We will check for skin irritation every time you switch back to your normal socket at the end of a session.

There are no known benefits for participating in this experiment. The prosthetic being developed is a prototype and will not be immediately available. You will be paid $8/hour for your participation in the study. If you chose to withdraw early from the study, you will still be paid for those sessions you attended. If any session is cut short because you feel discomfort or experience other physical difficulties, you will still be paid the same amount for the session.
The subject agrees to the following:

I am free at any time to seek further information regarding the experiment. Participation is voluntary and I am free to withdraw consent and discontinue participation at any time.

The subject will remain anonymous in all publications of the results of this experiment.

In the unlikely event of physical injury resulting from participation in this research, I understand that medical treatment will be available from the M.I.T. Medical Department, including first aid emergency treatment and follow-up care as needed, and that my insurance carrier may be billed for the cost of such treatment. However, no compensation can be provided for medical care apart from the foregoing. I further understand that making such medical treatment available; or providing it, does not imply that such injury is the investigator’s fault. I also understand that by my participation in this study I am not waiving any of my legal rights. Further information may be obtained by calling the Institute’s Insurance and Legal Affairs Office at 253-2822.

I understand that I may also contact the Chairman of the Committee on the Use of Humans as Experimental Subjects, M.I.T. 253-6787, if I feel I have been treated unfairly as a subject.

I have read the above consent document and understand the experiments described in the document. I agree to participate in the experiments as a subject.

The project investigators retain the right to cancel or postpone the experimental procedures at any time they see fit for medical or other causes.

Date: ______________

Subject’s Name: ______________________

Subject’s Signature: ____________________

Witness Name: _________________________

Witness Signature: _____________________
Appendix C

As a last note, a take home clinical trial was performed for a week with five subjects after the author’s research on this project was completed. Preliminary results from all subjects showed a small problem with the state transition to pre-swing in certain situations, which is under investigation. There was also some thought that a mechanical extension assist spring might be a useful addition. With these exception, the subjects were quite happy with the device, especially noting it’s added stance stability is making turns and climbing stairs.


20. Michael, John, CPO. Personal communication.

21. Kerrigan, Casey, M.D. Personal communication


23. Otto Bock, Inc. 3C100 Otto Bock C-Leg, Instructions for Use


