Modifying the MIT Sensorimotor Control Lab model of human balance and gait control for the addition of running

by

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ABSTRACT

This research continues the work begun by Sungho Jo and Steve G. Massaquoi on modeling human walking and upright balance. The model of human neurological control of balance and gait generation put forward by Jo and Massaquoi\cite{1} in “A model of cerebrocerebello-spinomuscular interaction in the sagittal control of human walking” and executed in MATLAB Simulink/SimMechanics.

This model has been used to determine the feed-forward command sequences for the generation of walking and running gaits. Furthermore, two feedback circuits controlling the center of mass relative to the swing leg and the composted leg angle of the simulated model were added. These provide a basis for a wider control of disturbances in order to implement running.

This work helps forward the long-term goals of the MIT Sensorimotor Control Group—creating a control model of the neurological circuitry responsible for governing human balance and locomotion and testing that model by using it to control a bipedal robot. The results of this research help to prove the validity of the cerebrocerebello-spinomuscular control model developed by Jo and Massaquoi and point positively towards the introduction of the running of the control model on a physical robot.

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I. INTRODUCTION
While a robot can lift a car off an assembly line much more easily than a team of human workers and an unmanned vehicle can explore Mars, where environmental conditions would kill a human, there are many human abilities that no robot can, at this point in time, reproduce with any degree of accuracy. One of these is bipedal locomotion—human walking. It is difficult to control the balance of a bipedal machine; legs, in effect, are two many-segmented inverted pendulum tied together. Before rushing ahead in the attempt to balance the machine, it is beneficial to understand the intricacies of how humans control themselves as human balance is certainly the best example of two-legged upright motion. The study of human upright balance and gait generation control is a challenge in and of itself, much less attempting to apply that control to a robot.

The entirety of this problem—finding the control algorithm governing human locomotion and then applying that model to a robot in order to prove its validity, is the challenging goal of the research being done by the Sensorimotor Control Group at MIT.

MOTIVATION FOR MODELING HUMAN MOTION IN ROBOTICS
A growing trend in the field of robotics is the replication of human dynamics, especially motion. Researchers are interested in creating humanoid robots for many reasons, including assisting people in every day tasks and do jobs impossible for humans. Perhaps most of all, researchers are interested in designing humanoid robots for the challenge of duplicating life in all its complexity and the possibility of improving on “human” design. When considering locomotion in particular, for example, a robot that maneuvers with the skill of a human would most likely be able to do any job currently done by a non-walking robot with improved results. A walking robot can cover rougher terrain, steeper slopes, handle stairs, or even jump obstacles. The most comprehensive answer, therefore, to Why attempt to build a human? is: for the pursuit of knowledge and for the practical benefits a humanoid robot offers.

CURRENT RESEARCH IN THE FIELD OF LEGGED ROBOTICS
There are many robots on the market or in the process of being built that move using legs and dynamic balancing. RHex, built by Boston Dynamics, has six legs jointed only at the hip which are capable of a full 360° of rotation. While this gives RHex incredible maneuverability over a wide variety of terrains (RHex handles mud, sand, rocks, thick underbrush and even underwater swimming), it is an artificial gait that works only for RHex’s unique body structure. RHex’s physical structure gives it an amazing ability to handle extreme terrains, but that same physical structure limits its functions and provides that RHex’s maneuverability remains unique to RHex.

Boston Dynamics comes closer to solving the problem of finding a control structure that could be applied across robots with BigDog. BigDog is a four legged robot, each leg having joints at the hip, knee, and ankle. This makes BigDog much more similar in physical structure to any four legged animal, and means that the theory of its control system for governing balance and gait generation could be applied across any other four legged robot, all of which have to deal with the problem of maintaining balance with multiple legs in the air and the need to switch gaits for various speeds.

While BigDog is a step in the right direction of modeling robotic control using similar life forms as a basis of research, walking or running with four legs does not present nearly the balance issues that two legs present. Honda’s robot Asimo and Sony’s robot Qrio are both humanoid robots that balance and move using two legs. Both robots use a control algorithm called zero moment point (ZMP). This keeps the point where the moments generated by angular momentum and ground contact forces are equal to zero within the zone of stability. ZMP is applicable across any two legged robot, but while fairly robust, it is computation intensive and works only on flat level terrain as the robot must compute every motion and carefully shift balance in order to provide a stable zone to move into. There is little of the free fall motion seen in dynamic walkers. Humans incorporate
both static calculation and dynamic movement in their gait, so while Asimo and Qrio are amazing achievements of balance there is still significant room for improvement in modeling human dynamics.

The robot Dexter, made by Anybots, is the first robot to walk using dynamic balancing, a combination of static calculation and balanced free-fall motion. It took its first steps in February of 2007. While Anybots has a long way to go to widen Dexter’s repertoire of gaits and ability to handle varied terrain, the fact that Dexter moves uses dynamic balancing is an incredible achievement.

**The Research Effort at the MIT Sensorimotor Control Group**

The theory behind the research of the MIT Sensorimotor Control Group is that in order to create the most effective and efficient control method for governing bipedal walking, engineers must study the existing successful model and inspiration for this effort—human locomotion. The goal of this research project is to determine how the human body controls upright balance and bipedal locomotion. While this research effort will produce a robot that moves by walking on two legs, this robot is a side product of successfully completing the goal of this project—modeling the neuro- and physiological control circuit used by humans to walk. While Asimo, Qrio, and now Dexter are testaments that bipedal walking and upright balance are possible, the terrain limitations and computation required are huge drawbacks to the ZMP control model used by Asimo and Qrio. Dexter has finally conquered the dynamic balance challenge, but while Dexter’s software learning program creates a final product that looks like human walking, how close the underlying structure mirrors the structure by which human walking is controlled is unknown. It is the goal of the MIT Sensorimotor Control Group to understand how humans control both balance and gait generation and the robot is the test device by which the results can be ratified. A robot capable of robust walking is the final goal of this research effort, but only as proof of the successful comprehension of the workings of the human body.

Understanding how the human body controls balance and gait generation has applications beyond creating a robust walking algorithm for robotics. In the medical field thus far, it has often been difficult to diagnose which specific parts of the brain or other elements governing walking are malfunctioning when a patient has a gait disorder. More detailed knowledge of how walking is generated and controlled will give rise to a better understanding of how walking and balancing can fail to be controlled. This will allow for better diagnoses of the causes of gait disorders and hopefully better cures.

As the control algorithm found by this research replicates human control circuitry, it provides a possible solution for patients suffering from gait disorders. One example is an active exoskeleton—an exoskeleton fitted to a patient’s legs and controlled with a model that replicates the human neurological control circuitry could sense (through, perhaps, the use of accelerometers or gyroscopes) when the patient’s movements became unstable and could actively assist the patient in correcting their gait.

**Scope of This Paper**

The long-term goals of the MIT Sensorimotor Control Group, with regards to this project, are as stated above: to create a control model of the neurological circuitry responsible for governing human balance and locomotion and to test that model by using it to control a bipedal robot. At this point in the research, a control theory modeling the cerebrocerebello-spinomuscular influences governing human balance and gait generation has been developed by Sungho Jo and Steve G. Massaquoi. The physical robot that this control model will be tested on is in the process of being developed. For the purposes of testing the control model while the robot is under construction, a MATLAB Simulink/SimMechanics model of the robot was created by Jo.

This paper presents work done using the MATLAB Simulink/SimMechanics simulation. Sungho Jo’s model was valid only for a control of basic walking. In order to extend that control to cover more unstable gaits, such as running, the control model and MATLAB simulation
developed by Jo was modified by adjusting the trunk pitch feedback loop, adding a forward phase to the basic gait cycle, and by adding two feedback circuits for (1) controlling the center of mass (COM) by adjusting the swing leg and for (2) controlling the relative angle of each leg to vertical.

Much of this work comes out of knowledge and experience gained through assisting with the construction of the bipedal robot “BOB” and designing and simulating human postural reflexes potentially useful in running. The former is summarized briefly and the latter which involved modifying MATLAB Simulink/SimMechanics simulation model is presented in greater depth.

The construction of BOB’s muscles, composed of series elastic elements (a weak and a strong spring coupled so that when the element is in tension only the force of the strong spring is seen, but when force is relaxed on the element, the weak spring provides the dominant force) provided an overview of basic musculature and how to model human physiology using mechanical components, an interesting counterpoint to the modeling of human physiology using MATLAB.

The MATLAB Simulink/SimMechanics modeling done in this research should be applicable in the physical domain of the robot as well as in simulation. While some gains will need to be adjusted when the model is run on the BOB (to account for muscles being composed of servo motors and series elastic elements, as well as having actual ground contact as opposed to the computed reaction forces) it can be assumed that the relationship between the parameters described in this paper remains fixed and only the scaling is affected. The theory of which muscles need to be activated at what point in time and to what degree relative to each other will remain constant.

II. BACKGROUND AND THEORY
The MIT Sensorimotor Control Group has created a model of the control system governing human upright walking. This model is based on the study of the human brain, specifically the spinal cord, cerebrum, and cerebellum, and is explained in detail in "A model of cerebrocerebellospiinomuscular interaction in the sagittal control of human walking" by Sungho Jo and Steve G. Massaquoi\textsuperscript{11}.

In this paper, Jo and Massaquoi divide human locomotor function into two categories, balance and gait generation. Balance control and gait generation are computed separately and then combined to form the command signal to the modeled muscles. The position of the body is then measured and fed back to the balance control and the gait generator. For a full explanation of the assumptions made by Jo and Massaquoi and the details of the model including how the neural pulses are generated and the balance is calculated, please see the full paper. However, a brief overview of the model is presented below.

Although a large number of systems potentially influence posture and gait, in an effort to create as simple a control system as possible Jo and Massaquoi focus on modeling the cerebrocerebellum and spinal cord as the main components governing balance and gait generation. Gait generation and balance control are handled separately within the model, and the control for each is based on the following assumptions:

One, that the basic components of a human’s gait are controlled by the brainstem and spinal cord. These generate a time-invariant, five-state train of electromyogram (EMG) pulses. This train of EMG pulses is then fed forward to activate muscle synergies controlling movement. This control structure is based out of research showing that EMG patterns are consistent across movement speeds, with only intensity and duration changing; which muscle synergies are activated at what time remains unchanged.

Two, that balance is governed by the cerebrum and the cerebellum. Body position is fed back to the cerebrocerebellar,
which controls the pitch of the trunk and the body center of mass. The cerebrocerebellar controls the center of mass by working from the stance leg and not by controlling the swing leg.

These assumptions provide the underlying structure for the walking control model.

III. PROCEDURE

In order to control gaits such as running and to develop a more robust control system, the MATLAB Simulink/SimMechanics file made by Sungho Jo and described in "A model of cerebrocerebello-spoionomuscular interaction in the sagittal control of human walking" was modified to include a forward swing phase, a feedback circuit for control of the swing leg, and a feedback circuit to control relative leg angle to the vertical. These changes are described in the following sections.

FEED-FORWARD CONTROL OF GAIT GENERATION: WALKING AND RUNNING

Jo's original MATLAB Simulink/SimMechanics model employed two different versions of the cerebrocerebellar trunk pitch control circuit. One control circuit was only valid for low gains, while the other circuit was valid for all gains. While the low gain control circuit was sufficient control under the low stresses of basic walking, it was judged to be insufficient under running conditions. The low gain control was removed from the model and the cerebrocerebellar trunk pitch control circuit described in "A model of cerebrocerebello-spoionomuscular interaction in the sagittal control of human walking" was used alone for control of the trunk pitch. This change kept the walking stable but allowed for the possibility of greater disturbances that would be present due to the high gains used in running.

Another small modification, in addition to the implementation of the cerebellar control loop, included changing \( k_x \), a constant involved in modeling ground contact. This was set to 1000 (some versions of Jo's model had this value while others were multiplied by 10). The ground contact model provided some implausible reaction forcers during the first step of the simulation. The initial starting position is chosen by the operator, and while Jo's initial position worked well for a walking gait, any attempt to run from this starting configuration produced reaction forces that either forced the figure to fly into the air or slip uncontrollably. For this reason, \( mus \) and \( muk \) are defined separately for walking and running, described later, and \( k_x \) was set to 1000. Any future work done with this model should take into account that the control structure being tested may very well work perfectly, but the combination of initial starting parameters and ground reaction modeling may be providing garbage values. This is mentioned again during discussion of the swing leg control.

The final basic modification to the MATLAB model was the addition of a fifth phase of the gait cycle. In his paper, Jo divides the human gait cycle into five phases—loading (LOA), regulation (REG), thrust (THR), retraction (RET), and forward swing (FOW). However, Jo found that activation of any muscles synergies during the FOW phase proved unnecessary for a basic walking simulation. He therefore included only LOA, RET, THR, and REG in the MATLAB Simulink/SimMechanics model (Jo and Massaquoi 14, 21). Activating the ankle dorsiflexor during the RET phase provided enough clearance between the foot and the ground so the model did not trip.

When the frequency and the gain of the neural pulses were increased in an attempt to speed the model and move towards running, it was found that activation of muscles only during the RET phase was insufficient to provide ground clearance. Therefore, a FOW phase was added to the model in order to lift the foot clear of the ground and to pull the swing leg sufficiently far ahead of the center of mass (COM).

The parameters defining the frequency of the forward phase, as well as the rest of the phases, are shown in Table 1. \( u_{ro} \) is the matrix defining the durations of each phase (Jo and Massaquoi 7).
\[ u(t) = \cos(2\pi \cdot \text{freq} \cdot \text{time} - \phi) - h_t \]

<table>
<thead>
<tr>
<th>( \phi )</th>
<th>( h_t )</th>
</tr>
</thead>
<tbody>
<tr>
<td>.5</td>
<td>.5275</td>
</tr>
<tr>
<td>.2</td>
<td>.705</td>
</tr>
<tr>
<td>.25</td>
<td>.95</td>
</tr>
</tbody>
</table>

Table 1: Parameters for the periodic pattern generation

\[
\text{MsynN} = \begin{bmatrix}
.3 & 0 & .4 & 0 & 0 \\
.38 & 0 & .64 & 0 & 0 \\
.8 & 0 & .8 & 0 \\
.7 & .7 & 0 & .3 \\
.0 & 0 & .4 & 0 \\
.0 & 0 & 0 & 0 \\
.0 & 0 & 0 & 0 \\
.0 & 0 & 0 & 0 \\
.0 & 0 & 0 & 0 \\
.0 & 0 & 0 & 0 \\
\end{bmatrix}
\]

Table 2: Stable walking synergies and amplitudes.

During loading (LOA) the gluteus maximus, which, as a hip extensor, forces the leg back, keeps the upper body from rotating forward as the heel strikes the ground.

In regulation (REG), the iliopsoas, vastus, and tibialis anterior muscles are activated. The iliopsoas is a hip flexor. It pulls the COM forward over the legs by pulling the trunk down to the knee. The vastus is a knee extensor. It forces the stance leg to support the body’s weight mid-stance. The tibialis anterior, an ankle dorsiflexor, brings the COM forward to help insure the correct stride length.

In thrust (THR), the vastus is still active from the REG phase in order to continue providing stance leg support. The soleus, an ankle plantarflexor, is activated in order to push the mass of the body forward and provide forward movement of the model.

Retraction (RET) activates the iliopsoas, biceps femoris short, and tibialis anterior. The iliopsoas activation forces the body to rotate forward about the hip and raises the leg. The biceps femoris short, a knee flexor, bends the knee to clear the foot from the ground. The tibialis anterior picks the toe up in order to avoid ground contact as well.

Finally, the tibialis anterior activation is continued through the forward (FOW) phase in

Figure 1: Active muscles during phases of walking gait

While additional explanation is given on page 8 of Jo’s “A model of cerebrocerebellospino-muscular interaction in the sagittal control of human walking,” the reasoning behind each muscle activation for each phase of walking (shown in Figure 1) is as follows:
in order to maintain ground clearance. While this was not originally included in Jo’s model, this added control during FOW swing was found to help prevent tripping.

These muscle synergies work well for relatively slow speeds, but several changes must be made in order to move more quickly. First, in order to move faster, the frequency of the neural pulses must be increased. This makes intuitive sense—to move faster, speed up the activation rate. Then, in order for the leg to move through the gait cycle at the increased pace, the gains of the neural pulses must be increased in order to increase muscle activation. This increase of neural pulse frequency and amplitude can go only so far, however. When the frequency of the neural activity becomes too great, the muscles can no longer respond quickly enough to follow the neural commands. The inability of muscles to respond quickly enough to the neural signals means that the speed of walking is limited. To move more quickly than a fast walk, the running gait is used.

There are a few key differences between a running gait and a walking gait. First, running is defined as having both legs off the ground at some point during the gait cycle. This is never true during normal walking. Second, while in walking the swing leg straightens before landing, in running the swing leg (which straightened in order to push off the ground) bends as it contacts the ground. This helps absorb the shock of landing and preload the leg for the next push off.

The walking model described above, with the FOW phase and the trunk pitch controller as mentioned, was modified for basic running. The frequency and intensity of the pulse train was increased in order to move the figure forward more quickly and the muscle activation matrix was modified. The muscle synergies activated during various phases of the gait cycle were changed in order to better handle the increased frequency and gains. The modified settings are shown in Table 3 and Figure 2.

![Figure 2: Active muscles during phases of running gait.](image)

**Table 3: Experimental values and gains for running.**

An explanation of each muscle activation for each phase is as follows:

As in the walking case, the gluteus maximus is active during LOA to keep the upper body from rotating too far forward as the heel strikes the ground. However, to help absorb the force of the landing, the biceps femoris short is activated to bend the knee.
In REG, the iliopsoas and biceps femoris short muscles are activated. The iliopsoas pulls the COM forward over the legs by pulling the trunk down to the knee while the biceps femoris short remains bent from the LOA phase to keep the weight low. This is directly opposite the walking case, where the vastus straightened the stance leg out. Also, the tibialis anterior is not activated here as it was in walking, but as these values are far from final it may be beneficial to later activate these muscles if the COM needs to be brought forward to insure the correct stride length.

In THR, the vastus and the soleus become active just as in walking, providing thrust to propel the body forward. Here, the magnitude of the vastus and soleus activation is greatly increased, especially the vastus.

The RET phase activates the iliopsoas, biceps femoris short, and tibialis anterior as in walking. The iliopsoas activation forces the body to rotate forward about the hip and raises the leg, the biceps femoris short bends the knee to clear the foot from the ground, and the tibialis anterior picks the toe up in order to avoid tripping.

The iliopsoas, biceps femoris short, and tibialis anterior remain activated through the FOW phase in order to maintain ground clearance and pull the leg forward far enough to remain in the air until the stance leg pushes off.

These settings result in body movements that begin to approach a running gait. However, the model is extremely unstable with these values; the COM control and trunk pitch control are unable to regulate the accumulated errors generated by this series of feed-forward pulses, and the model stays upright for only one to two steps. However, as the motion of these muscle synergies during these phases of the gait cycle do begin to approximate the desired leg motions, these values are shown here as they provide a good starting point for further research.

**FEEDBACK CONTROL: SWING LEG CONTROL**

**OF COM AND RELATIVE LEG ANGLE CONTROL**

Jo’s control model governs balance by regulating the position of the center of mass (COM) relative to the stance foot. This works well in basic walking, where one foot is on the ground at all times. However, during running both feet must be off the ground and so the model will be unable to regulate it’s COM.

To overcome this limitation and maintain balance when running, a feedback control circuit controlling the COM relative to the swing leg was added to the model. This circuit first checks to make sure that a leg is swinging by making sure that the leg is not in contact with the ground. If the leg is in swing, then the error between the current COM position relative to the stance foot and the desired step length divided by two is calculated (Equation 1). The figure is stable when the COM rests in between the two feet. This is why the desired step length (desw) divided by two is used as a reference signal—when the model is in a stable position, the COM will be directly in between the two feet, located desw apart.

\[
\text{error} = \left(\frac{\text{desw}}{2}\right) - (x \text{ position COM relative to stance foot})
\]

The error is scaled by a gain and multiplied across an activation matrix activating either the iliopsoas or the gluteus maximus of the swing leg. If the error is positive, the COM is not far enough forward and so the iliopsoas of the swing leg is activated, bringing the swing leg farther forward and causing the COM to move closer to desw/2. If the error is negative, the COM has gone beyond desw/2 and so the gluteus maximus of the swing leg is activated to draw the swing leg back. This muscle activation matrix is then multiplied by \(u_{p}\), the matrix defining the durations of each phase of the gait cycle, and combined with the feed-forward signals from the neural pulse generators.
An example of this control circuit used on an unstable system is shown in Figure 4. This walking model had the inhibition gains for the ankle, knee, and hip set to 0.8, causing it to stumble after roughly 5 steps. Using the swing leg control algorithm with the gain set to 1, the model walks a step or so further. This trial was chosen as an example as the frames of the model animation show clearly how the swing leg control is affecting the model.
Figure 4: Example of swing leg control on an unstable model.

Looking at pictures 15 through 20, for example, the influence of the swing leg control is easily seen. The trunk is tilted back, and so the swing leg is lifted high above the waist. The model tilts forward, and the swing leg is returned to the ground.

Pictures 36 through 46 show how the swing leg control model fails in this instance. The swing leg is lifted high to counteract the trunk's lean backwards, but the stance leg also compensates and so the swing leg comes down abruptly (39-40) causing reaction forces that the model cannot handle.
The swing control algorithm, when used on an already stable system such as the walking model described earlier in this paper, makes the system unsteady very quickly unless the gain is kept below .02. As the gain of the swing leg control is increased, the model brings its swing leg forward too high in an attempt to bring the COM forward to $d_{sw}/2$ and has no time to return to the ground before the stance leg pushes off. Also, the failure seen in Figure 4 may very well be due entirely to the ground contact model. Simply bringing the model’s heel down sharply should no generate enough force to catapult the model into the air. The control model is doing exactly what it is built to do; other forces that the swing control are working to destabilize the system.

A gain of 1 in the swing control model allows the swing control to positively affect an unstable system, yet drives a stable system unstable. This, however, may not be the fault of the swing control model. The swing control does help bring the COM into a more stable position no matter which system it is used on; the instability lies in the fact that the feed-forward pulses of the rest of the gait are not modified to account for the change in the swing leg. By itself, therefore, the swing leg control may not be applicable across multiple models. Used in combination with the composite leg angle control it may provide a more robust method of support.

The composite leg angle control method constrains the composite angles formed by the legs to be equal and opposite to each other. The composite leg is the ray formed by connecting the hip joint with the ankle joint—see Figure 5 for a drawing of this.

In walking and running, stability is provided by keeping the COM as centered as much as possible between the two legs, meaning that the angles formed by the composite legs are equal and opposite to each other—one is forward (positive) while the other is backwards (negative). While COM control based off of the stance leg and the swing leg tends to force this opposite sign, equal magnitude relationship between the composite legs, it does not do so directly. Therefore, the addition of a control circuit explicitly enforcing the equal and opposite relationship between the composite legs will help prevent the type of instabilities seen under the swing control model of stability. For example, where the swing leg was raised very high in order to pull the COM forward, the composite angle of the swing leg was much greater than that of the stance leg. Had the composite angle of the stance leg widened in response to the raising of the swing leg, three improvements would have occurred immediately: (1) the COM would have moved forward into a stable region much more quickly than with the swing leg control alone, (2) the swing leg would not have needed to travel as far towards the trunk, out of the way of its normal gait pattern, and (3) the swing leg would also not have had as far a distance to travel to return to its normal path, meaning that the swing leg would have been in place to receive the transferred weight off of the stance foot.

![Figure 5: Model of parameters used in composite leg angle control.](image)

A composite leg angle control was implemented in an effort to achieve the before mentioned improvements in the swing leg control. The composite angle is computed by looking at the geometry of the leg, shown in Figure 6: **Geometry of composite leg.** The composite angle is calculated according to Equations 2 through 9. The sign of the composite leg angle
is positive when the leg is forward of vertical and negative when the leg is behind vertical.

\[ \text{kneepos}_x = \text{thigh} \times \sin(\text{hip} - \Theta_{tr}) \]
\[ \text{kneepos}_y = \text{thigh} \times \cos(\text{hip} - \Theta_{tr}) \]
\[ \text{anklepos}_x = \text{shin} \times \sin(\text{knee}) \]
\[ \text{anklepos}_y = \text{shin} \times \cos(\text{knee}) \]
\[ \text{composite}_x = \text{kneepos}_x + \text{anklepos}_x \]
\[ \text{composite}_y = \text{kneepos}_y + \text{anklepos}_y \]
\[ \text{composite\_length} = \sqrt{\left(\text{composite}_x\right)^2 + \left(\text{composite}_y\right)^2} \]
\[ \text{composite\_angle} = \sin\left(\frac{\text{composite}_x}{\text{composite\_length}}\right) \]

**Figure 6:** Geometry of composite leg.

The control of the composite leg angle was calculated by adding the right leg composite angle to the left leg composite angle and comparing the result to zero. The magnitude and signs of the right and left composite angles should be equal and opposite for stability, and will therefore sum to zero. If the result of this addition was positive, than the gluteus maximus muscles of both legs were activated. The result is positive under two conditions: (1) the forward leg angle is greater in magnitude than the behind leg angle or (2) both legs are forward. In either case activating the gluteus maximus of each leg helps stabilize the system. If case (1), then the front leg is pulled back to decrease the composite leg angle while the back leg is pulled back to increase the back leg angle. If case (2), both legs are pulled back together closer to center.

If the result of adding the right composite leg angle to the left composite leg angle was negative, than the iliopsoas muscles of both legs are activated. The result of the angle addition is negative under two conditions: (1) the rear leg angle is greater in magnitude than the front leg angle or (2) both legs are behind vertical. In either case activating the iliopsoases of each leg helps stabilize the system. If case (1), then the front leg is pulled forward, increasing the forward angle, while the back leg is pulled forward, decreasing the back angle. If case (2), both legs are pulled forward together closer to center.

**IV. RESULTS AND DISCUSSION**

The cerebrocerebellar long loop feedback control of the COM and the addition of a FOW phase are important contributions to the basic walking model. Their addition preserves the stability of the model and provides a basis for the addition of parameters needed for gaits more disturbing to stability such as running.

The fact that the swing leg control does help to somewhat stabilize unstable systems shows the promise of this control method. While the swing leg control may be a good idea overall, this control circuit needs to be adjusted to provide stability over a wider range of movements. While some of this stability will come from choosing a gain appropriate for the situation, the addition of the composite angle control helps to improve balance as it affects the stance leg to the same degree as the swing leg. This helps the stance leg “keep up” with the movement of the swing leg control circuit. With this help, the swing leg does not need to move as quickly or as far in its attempt to balance the COM, as the shift of the stance leg as it attempts to keep the composite leg angle of the stance leg equal to that of the swing leg helps move the COM further towards stability.
V. SUGGESTIONS FOR FURTHER RESEARCH
While an important start has been made on balancing the model under conditions such as running, much has yet to be done to find the optimal gains and parameters that make the swing leg control and the composite leg angle control capable of controlling a model under stable and unstable circumstances. While the swing leg control works to control the model under some conditions, such as when the inhibitory gains are lowered to cause instability, swing leg control fails to assist in improving an already stable or semi-stable system. Rather, it causes the system to fail sooner than it otherwise would have or fail at all when it would have remained stable indefinitely.

This version of the composite leg angle control activates only the hip flexor and extensor in order to move adjust the composite angles. The knee as well as the hip, however, also contributes to the makeup of the composite leg and composite leg angle. Expanding this control model to include control of the knee would help stabilize the model, as the legs would not need to be moved so far out of the gait path in order to keep the angles complimentary.

Once the control circuits for swing leg control and composite leg angle control have been satisfactorily adjusted to improve (or at the least not upset) stable walking, the feed-forward pulses for running should be revisited. Working off of the pulses presented here, the optimal muscle synergies to be activated during each phase of the running gait as well as the magnitude of the activation pulses should be determined. This should provide for a stable and robust running gait.

These actions will go a long way towards developing a complete model of the human neuro-muscular control of bipedal locomotion. The final and most exciting step is, of course, to test the simulation results on a physical system—a bipedal robot. Testing the model on a physical system will bring its own host of challenges, but the opportunity for the amazing achievement of building a dynamic humanoid robot.

REFERENCES

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