

Hip Function Characterization in the Sagittal Plane with Varying Gait Speed

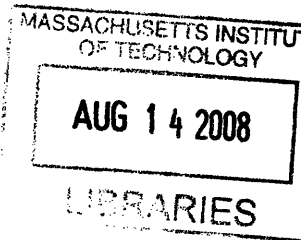
by

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# Hip Function Characterization in the Sagittal Plane with Varying Gait Speed

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## ABSTRACT

The function of the human hip joint during the stance phase of walking can be characterized with a configuration of simple mechanical elements. This combination of elements is capable of providing general hip behavior in the sagittal plane. Data was collected from two healthy, young subjects who walked at slow, normal and fast gait speeds.

The hip can be modeled with a torque actuator and two independent, linear torsional springs, which are activated at different times during the stance phase of gait. The activation times consistently identify gait cycle events across all three gait speeds. The first spring operates during the single limb stance of the gait cycle. The second spring is actuated during second double support, in the pre-swing phase. The springs effectively reduce the amount of work an unaccompanied torque actuator would have to exert in order to reproduce the hip gait pattern.

Thesis Supervisor: Hugh M. Herr

Title: Instructor, Harvard-MIT Division of Health Sciences and Technology

To my family

Mom, Dad, and Berenice

Because I would never have been able to make it this far without them

Thank you for your support

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## Section 1

# Introduction

### 1.1 Motivation and Approach

The goal of this study is to characterize human hip function in the sagittal plane during normal walking. This simple model of the hip system can be useful in the development of an artificial hip. Standard hip performance can be derived from such a model and thus provide a way to compare the performance of an artificial hip to that of a biological hip. During development of complex biomechanical systems such as robotic prostheses, it is imperative to take into account performance and design considerations. Since compromises have to be made throughout development, it is important to identify these decisions early in the design. Using a simple model to achieve a precise system early in the design process can eliminate extraneous and costly iterative prototypes. Furthermore, a simple hip model can be utilized in developing corrective orthotic devices as well, since the model provides a quantitative behavior to correct abnormal pathological gait. Finally, the modeling of hip behavior also demonstrates variation in the model's parameters from one gait speed to another. Understanding when and to what extent these parameters change, insight can be gained about the control strategy that governs normal function (Palmer 2002, et. al.).

A simple model configuration and its parameters can be derived using gait data collected from normal subjects. The comprising elements must be simple mechanical elements which are capable of imitating hip value given kinematic inputs such as position and velocity. Furthermore, gait speed variations are also taken into account in order to develop a model capable of adjusting to various gait speeds.

## Section 2

# Methods

### 2.1 Data Collection and Processing

The subject for this research included two healthy, male volunteers. The subjects' age, mass, and anthropometric data are shown in Table 2.1.

**Table 2.1** Subject descriptions

| Subject Label | Sex  | Age<br>[yrs] | Mass<br>[kg] | Height<br>[m] | Mean Leg Length<br>[m] |
|---------------|------|--------------|--------------|---------------|------------------------|
| DMG           | M    | 26           | 70           | 1.83          | 1.02                   |
| FJI           | M    | 27           | 84.5         | 1.82          | 0.97                   |
|               | Mean | 26.5         | 77.3         | 1.8           | 1.00                   |
|               | SD   | 0.7          | 10.3         | 0.0           | 0.04                   |

This data were collected during an independent study (Riley et. al. 2001) and has been used in a previous project which involved the biomechanic modeling of the ankle (Palmer et. al. 2002). The following description is a very brief summary of the procedures and set up used for collecting the sagittal data. Subjects walked barefoot at their self-selected normal speed on a 10 meter walkway. They were then asked to walk “faster then you would normally walk” for several trials, as well as “slower then you would normally walk”. Kinematic and kinetic data were acquired for both lower limbs via a six-camera VICON 512 system (Oxford Metrics, Oxford, UK) and two AMTI force plates (AMTI, Newton, MA). The data were processed at 120Hz with VICON



BodyBuilder (Oxford Metrics, Oxford, UK) using the lower limb standard model included with the software. The data derived using the standard BodyBuilder model, including hip angular position and hip torque, were then analyzed using MATLAB (MathWorks, Natick, MA) (Palmer 2002, et. al).

The following sign conventions were used: positive hip position for hip extension and positive hip torque for extension torque Zero hip position was assigned to be the moment in time at which the foot is perpendicular to the shank (Rose 2005).

Hip angular velocity was calculated by filtering the ankle position data and then numerically differentiating. Although it is customary to apply a zero-lag, fourth order, low-pass Butterworth filter (Winter 1990, Palmer 2002), this filter caused an over-idealization of the data which then proved to be problematic. Instead, numerical integration was performed on the original angular position data and filtered using a least-squares, Savitzky Golay filter.

Hip power was found as the product of the hip torque and hip velocity magnitudes in the sagittal plane. All other out-of-plane components of angular velocity or torque were not included in the overall power calculation. Furthermore, work done during a certain time period was found by integrating the power versus time curve.

The gait speed for each trial was taken to be the slope of the line given by

$$x = vt + x_0 \quad (2.1)$$

Where  $v$  is the slope,  $t$  is time,  $x$  is the absolute position of a point fixed in the pelvis, and along an axis parallel to the forward motion direction, and  $x_0$  is the position at time equal zero. Slope  $v$  was easily found with linear regression. (Palmer 2002, et. al)

## 2.2 Characterizing Hip Function

The methodology of this research is predominantly based on a project which characterized the ankle complex in the sagittal plane, and during various gait speeds (Palmer 2002, et. al.). The approach can be described as “the black box” method. The ankle joint, or in this case the hip joint is characterized by “a black box” containing an arrangement of simple mechanical components. The goal is to identify the contents of the “black box” in order to model hip behavior during gait. Using experimental kinematic and kinetic data (hip angular position  $\theta$ , hip torque  $\tau$ , hip power  $P$ ) and understanding the behavior of simple mechanical elements, the contents of the box can be identified.

Hip angular position and angular velocity were considered as inputs to the mystery system. Hip torque and power were set as outputs. Position/torque and velocity/torque relationships were used to derive the appropriate combination of mechanical components. It is necessary to understand that this approach ignores the individual functions of the muscles, tendons and bones which comprise the hip joint. Instead, the model quantifies the function of the hip complex as a whole entity.

In order to identify the contents of the black box model, the possible mechanical elements must be defined. Three mechanical elements were considered: torsional springs, torsional dampers, and torque actuators (motors). The springs and dampers are passive elements and considered to characterize hip function when power is negative. Torque actuators are active elements and considered during positive hip power. Actuators are also considered when the amount of positive work exceeds the stored energy in the system.

When considering a mechanical arrangement, it is necessary to separate spring-like behavior to that of damper-like behavior. The two can be told apart when no correlated variation exist between angular position and angular velocity. During these time periods, it is necessary to calculate the time difference between the occurrence of the maximum torque magnitude and that of maximum angle position magnitude. A small time difference, or phase difference, identifies spring-like behavior in a given time segment.

Spring elements can be either linear or non-linear. A linear spring exhibits a direct correlation between torque and angular position as seen in Equation 2.2.

$$\tau = K\theta + \tau_0 \quad (2.2)$$

The spring is characterized by the spring stiffness  $K$  and the initial torque,  $\tau_0$ . These parameters can be found using linear regression of the torque versus angular position data. If the data does not fit this model, the spring is considered non-linear.

### **2.3 Proposing a Model and Characteristic Parameters**

A configuration of elements can be proposed and then tested with the black box method by inputting angular position and velocity into the model. An educated guess can be derived by observing the general behavior of the hip joint. The hip power profile demonstrates two potential spring-like sections, where energy is stored, then immediately dissipated. This observation leads to a possible arrangement of two clutch activated springs and a torque actuator. These springs are independent of one another (neither in series nor in parallel), and activated at critical times,  $T_1$  and  $T_2$  respectively. The stiffness constants,  $K_1$  and  $K_2$  may be constant or velocity dependent.

The proposed configuration initiates spring 1 at  $T_1$ , and expires the element once all of its stored energy is released. Soon after, spring 2 is activated at  $T_2$  until that spring's energy is dissipated. The torque actuator performs the necessary work (either positive or negative) to replicate the general hip behavior.

The four characteristic parameters can be found with an iterative procedure using MATLAB. The process evaluates hip data at a time segment which begins at  $T_1$ . First, a  $\tau$  vs.  $\theta$  linear regression specifies the spring parameters given by Equation 2.2. The resultant modeled power can then be derived from the modeled torque. Mechanical power is defined as

$$P = \tau\omega \quad (2.3)$$

Where  $\omega$  is the hip angular velocity and  $\tau$  is hip joint torque. The modeled power can then be compared to the biological power and evaluated for “goodness”. The process repeats until the clutch times,  $T_1$  and  $T_2$  and spring stiffness'  $K_1$  and  $K_2$  provide the least error between modeled power and biological power.

## Section 3

# Results

### 3.1 Gait Parameters

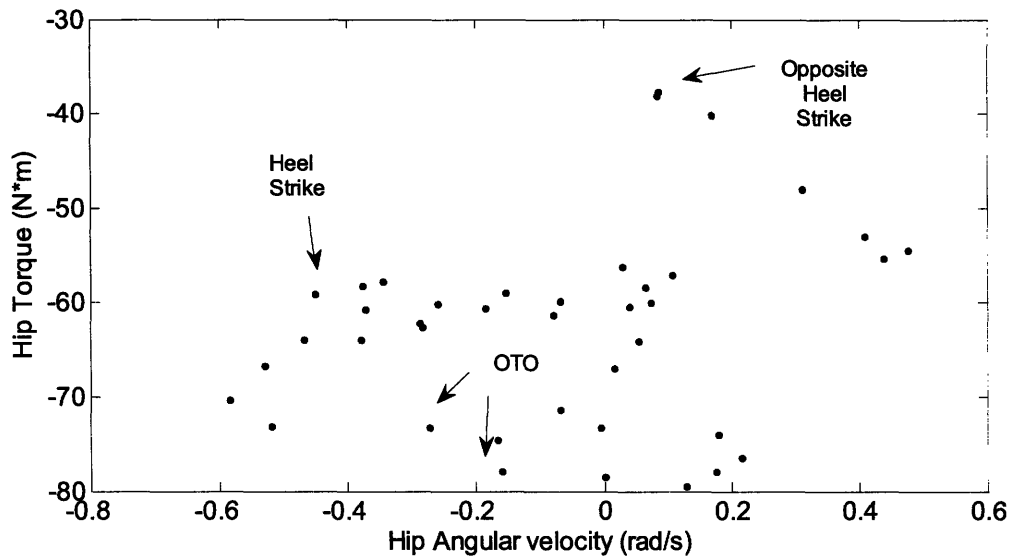
Time and distance gait parameters for the hip were calculated for each subject at slow, normal and fast gait speeds. (Table 3.1)

**Table 3.1** The subjects' mean (SD) time and distance gait parameters

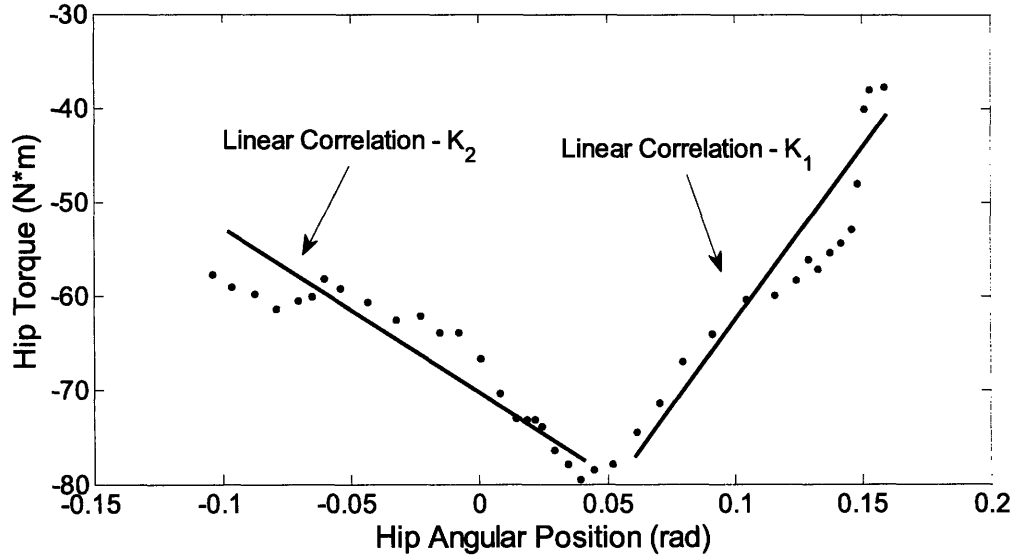
| Subject Label | Self-Selected Speed | # of Trials | Gait Speed [m/s] | Stride Length [m] | Duration of Gait Cycle [ms] | Duration of Stance [%GC] |
|---------------|---------------------|-------------|------------------|-------------------|-----------------------------|--------------------------|
| DMG           | Slow                | 6           | 0.77(0.019)      | 1.090(0.029)      | 1412(23)                    | 65(1.4)                  |
|               | Normal              | 5           | 1.30(0.060)      | 1.437(0.045)      | 1107(20)                    | 62(0.7)                  |
|               | Fast                | 8           | 1.69(0.038)      | 1.689(0.034)      | 999(17)                     | 61(1.0)                  |
| FJI           | Slow                | 6           | 0.87(0.022)      | 1.163(0.044)      | 1351(50)                    | 67(1.4)                  |
|               | Normal              | 7           | 1.21(0.016)      | 1.354(0.014)      | 1119(20)                    | 65(1.4)                  |
|               | Fast                | 5           | 1.69(0.063)      | 1.607(0.035)      | 950(19)                     | 62(0.6)                  |

### 3.2 Validating the Proposed Configuration

As discussed in the methodology, both damper-like and spring-like elements can be considered during the stance phase of the gait cycle. The proposed black box configuration only uses springs as passive elements and this assumption can be validated. Scatter plots of the hip torque versus hip angular velocity (Figure 3.1) in the stance phase region, do not suggest a direct correlation between hip torque and hip angular velocity. However, torque versus angular position scatter plots demonstrate a linear correlation which validates the assumption of linear spring-like behavior. (Figure 3.2)



**Figure 3.1** Scatter plot of hip torque versus hip angular velocity during early stance phase. Data shown are from a single DMG trial at self-selected normal speed. Data are equally spaced in time.



**Figure 3.2** Results of simple linear regression of hip torque on hip angular position during early stance phase for a single, normal speed trial. Data shown are from the same trial as Figure 3.1. Data points are equally spaced in time.

### 3.3 Clutch Time

The clutch times  $T_1$  and  $T_2$  were found for all trials and noted in Table 3.2.

Clutch times tended to remain in close proximity (standard deviation less than 0.05 seconds), within gait speed groups. Note that average values of  $T_1$  and  $T_2$  are trivial in this situation due to the variance between trials. Time landmarks must be evaluated with respect to its own time period.

**Table 3.2** Results of clutch time optimization for subjects during the stance phase.

| Gait Speed Group | DMG     |           |           | Gait Speed Group | FJI     |           |           |
|------------------|---------|-----------|-----------|------------------|---------|-----------|-----------|
|                  | Trial # | $T_1$ (s) | $T_2$ (s) |                  | Trial # | $T_1$ (s) | $T_2$ (s) |
| Slow             | 14      | 0.117     | 0.342     | Slow             | 14      | 0.117     | 0.342     |
|                  | 15      | 0.108     | 0.342     |                  | 15      | 0.108     | 0.342     |
|                  | 16      | 0.05      | 0.392     |                  | 16      | 0.125     | 0.408     |
|                  | 17      | 0.167     | 0.417     |                  | 17      | 0.108     | 0.467     |
|                  | 18      | 0.125     | 0.408     |                  | 18      | 0.119     | 0.392     |
|                  | 19      | 0.108     | 0.467     |                  | 19      | 0.167     | 0.417     |
| Normal           | 1       | 0.108     | 0.267     | Normal           | 1       | 0.108     | 0.267     |
|                  | 2       | 0.108     | 0.308     |                  | 2       | 0.117     | 0.333     |
|                  | 3       | 0.108     | 0.325     |                  | 3       | 0.125     | 0.317     |
|                  | 4       | 0.125     | 0.317     |                  | 4       | 0.108     | 0.325     |
|                  | 5       | 0.117     | 0.333     |                  | 5       | 0.108     | 0.308     |
| Fast             | 6       | 0.092     | 0.308     | Fast             | 6       | 0.112     | 0.308     |
|                  | 7       | 0.1       | 0.375     |                  | 7       | 0.108     | 0.4       |
|                  | 8       | 0.1       | 0.333     |                  | 8       | 0.1       | 0.333     |
|                  | 9       | 0.1       | 0.417     |                  | 9       | 0.083     | 0.4       |
|                  | 10      | 0.1       | 0.4       |                  | 10      | 0.1       | 0.4       |
|                  | 11      | 0.091     | 0.4       |                  | 11      | 0.091     | 0.4       |
|                  | 12      | 0.083     | 0.4       |                  | 12      | 0.1       | 0.417     |
|                  | 13      | 0.091     | 0.392     |                  | 13      | 0.091     | 0.392     |

### 3.4 Spring Stiffness

Furthermore, spring stiffness values,  $K_1$  and  $K_2$  were derived simultaneously with clutch times (Table 3.2). As gait speed increases,  $K_1$  values increase to provide a stiffer spring. Conversely,  $K_2$  values decrease as gait speed increases. The negative sign

accounts for a clockwise change in angular position during the second half of the stance phase.

**Table 3.3** Average spring stiffness values found via linear regression of hip torque on hip angular position. Iterative process minimized error in order to optimize  $K_1$  and  $K_2$ . All values are the mean (SD) for each group.

| Gait Speed Group | DMG             |                  | FJI             |                 |
|------------------|-----------------|------------------|-----------------|-----------------|
|                  | $K_1$ [N*m/rad] | $K_2$ [N*m/rad]  | $K_1$ [N*m/rad] | $K_2$ [N*m/rad] |
| Slow             | 20.239 (32.172) | -238.820 (51.41) | 34.76(20.22)    | -167.74(51.42)  |
| Normal           | 75.046 (13.947) | -90.082 (58.41)  | 41.24(39.12)    | -105.98(58.42)  |
| Fast             | 65.476 (20.816) | -25.379 (17.17)  | 68.21(32.43)    | -48.30(40.47)   |

### 3.5 Work Evaluation

Once the four characteristic parameters,  $T_1$ ,  $K_1$ ,  $T_2$ ,  $K_2$ , were optimized, the effectiveness of the model must be evaluated by calculating the power contribution provided by the springs. As earlier mentioned a torque actuator, or motor, provides additional torque required to generate the hip power profile. The residuals for modeled and experimental power demonstrate to additional power the motor must provide during the spring-modeled periods (Figure 3.3). Integrating the power residuals and the biological power during the specified time periods provides the work performed by the active and passive elements (Table 3.4). Furthermore, the percentage of total work savings provided by the springs can be calculated as

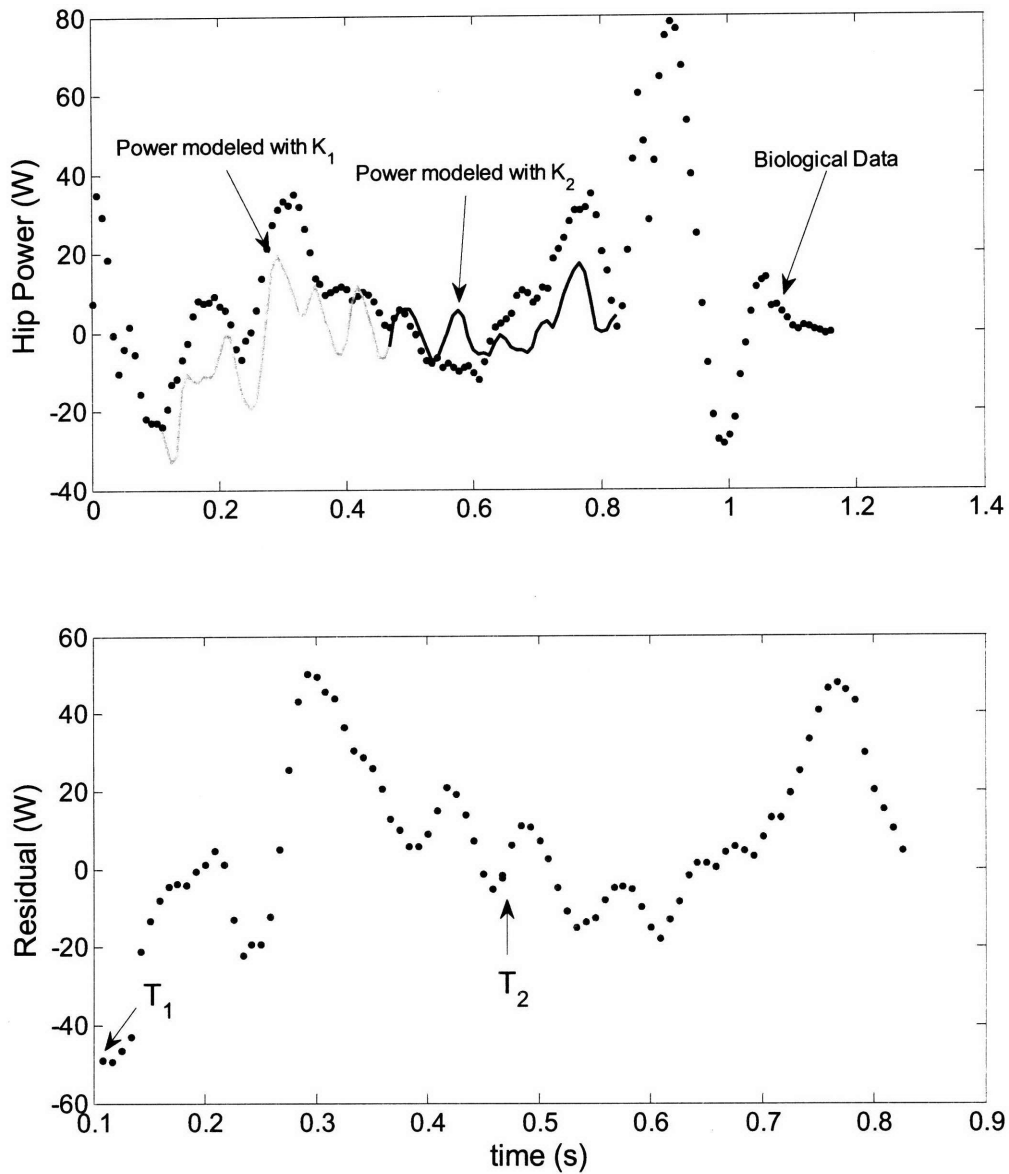


$$\%Savings = \frac{W_{S1} + W_{S2}}{W_{total}} \times 100 \quad (3.5)$$

Where  $W_{total}$  is the work output of both springs and the motor. On average, the springs provided 34% savings at normal walking speed, 50% savings during fast walking, and 65% savings at slow speeds.

**Table 3.4** Results from numerically integrating power versus time data.  $W_{S1}$  and  $W_{S2}$  denote work done by the spring model.  $W_{M1}$  and  $W_{M2}$  are work quantities provided by the torque actuator. Percent work saved by springs is calculated with Equation (3.5). Values given are the mean (SD) for each speed group.

| Gait Speed | $W_{S1}$ [J]   | $W_{M1}$ [J]   | $W_{S2}$ [J]  | $W_{M2}$ [J]   | $W_{S1} + W_{S2}$ [J] | $W_{M1} + W_{M2}$ [J] | %Work Saved by Springs |
|------------|----------------|----------------|---------------|----------------|-----------------------|-----------------------|------------------------|
| Normal     | 10.711 (1.208) | 7.133 (0.610)  | 2.644 (0.489) | -0.092 (0.212) | 13.355                | 7.041                 | 34.5                   |
| Fast       | 7.874 (5.960)  | 10.493 (3.063) | 3.524 (1.433) | 0.389 (2.515)  | 11.398                | 10.882                | 49.9                   |
| Slow       | 2.513 (1.451)  | 4.711 (3.056)  | 0.761 (0.829) | 1.957 (0.498)  | 3.274                 | 6.667                 | 65.3                   |



**Figure 3.3** Subject DMG, biological power data and modeled hip power plotted against time. Residuals for modeled power profile demonstrate the additional power necessary from a torque actuator to achieve hip behavior.

## Section 4

# Discussion

### 4.1 Subject Normality

The data provided by the subjects of a study can be classified as “normal” if they meet certain gait parameters established in previous reports. The ankle characterizing study qualified its subjects as normal. Since this study utilizes hip data from two of the same subjects, taken during the same sessions as the ankle study, normality is established by association.

### 4.2 Spring Activation Times

It was mentioned earlier that clutch times should be evaluated within their own gait cycles. The spring activation times resulted as characteristic time landmarks within their gait cycles. Clutch times  $T_1$  consistently marked the opposite toe off (OTO) event which typically occurs at about 12% of a normal walking cycle. The moment became later in the gait cycle as gait speed decreased because the initial double support period lasts for a longer amount of time during slower speeds. Accordingly,  $T_1$  occurs earlier in fast gait speed trials.

The second spring activation time  $T_2$ , occurs at opposite heel strike, which occurs at about 50% of a normal walking cycle. The result was consistent throughout all trials, and as shifted accordingly with changes in gait speed. These clutch times characterize the initial spring time period which occurs during the single limb stance (SLS) of the cycle. The second spring is used up by the time the cycle reaches toe off (TO) and the

swing phase begins. This means that the second spring spans the second double support (SDS) phase.

The proposed model is based on the observation of stored and then dissipating energy in the power profile. The spring activation times validate this observation and demonstrate predominant spring behavior during the stance phase.

### 4.3 Spring Stiffness

$K_1$  and  $K_2$  further characterize the function of each of the springs during their corresponding gait cycle period.  $K_1$  is primarily positive during SLS. The positive sign denotes the hip extension which occurs throughout mid-stance and the hip hyperextension during terminal stance. As gait speed increases, the hip must resist the excessive hyperextension to avoid injury. The increase in spring stiffness during SLS is a precautionary measure which adapts with speed.

$K_2$  values are negative throughout SDS due to the reversal in hip behavior. It is during this phase that weight is transferred from one limb to the other and the hip rapidly flexes back to neutral position. The rapid reversal may account for  $K_2$  greater magnitudes than  $K_1$  magnitudes. The  $K_2$  values demonstrate a negative correlation with gait speed. As gait speed increases, the leg acquires the necessary momentum to reverse the hip's behavior from extension to flexion. The lower stiffness allows for a quick reversal and does not impede the behavior necessary for weight transfer.

#### **4.4 Model Effectiveness and Concluding Remarks**

The percentage of energy savings provided by the springs demonstrates a promising model for the hip joint complex. Fast and slow speed group models were capable of achieving more than 50% work savings in torque actuator output. Both fast and slow gait speed groups obtained better overall models than normal speed groups, since modeled data fit the biological data more closely. Large discrepancies between model output and biological output call for additional work required from the assisting torque actuator. Future work would involve a greater focus on modeling with an additional gait speed dependency parameter.

This model can be used to develop a robotic hip capable of replicating general hip behavior using time activated torsional springs and a torque actuator to supply additional power. A feedback system positioned at the torso would provide further information to regulate motor output and effectively behave like a biological hip. A design such as this could potentially be developed into an Orthotics device which minimizes the pain many individuals experience when forced to overcompensate for an inadequate hip joint.

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