INTRODUCTION
A major goal in prosthetic knee design is to use mechanical and electrical components to restore the lost musculotendon function of above-knee amputees, allowing them to walk with able-bodied kinematics. Proper selection of mechanical components is critical for low-cost, passive knees, such as those used in low-income settings and developing countries. Previous studies have selected mechanical components by dividing the gait cycle into multiple phases, defining a mechanical model during each phase, and optimizing the coefficients of the model to reproduce the moment-time relationship corresponding to able-bodied kinematics (Sup, 2008; Martinez-Villalpando, 2009). However, these studies targeted the moment-time relationship required for able-bodied humans to walk with able-bodied kinematics, rather than that required for above-knee amputees.

The present study calculates the moment-time relationship for above-knee amputees to walk with able-bodied kinematics and optimizes coefficients of a mechanical model to reproduce this relationship. The optimized coefficients are computed for a wide range of prosthesis mass configurations.

METHOD
A 3-segment link-segment model of a prosthetic leg consisting of an upper leg, lower leg, and foot was defined, and 2-dimensional inverse dynamics was performed. Normative gait kinematics (Winter, 1991) were prescribed. Various masses were prescribed to each segment of the model to represent a range of prosthesis mass configurations, and the gravitational force on each segment was prescribed in proportion to its mass.

Since the net force acting on the body – i.e., gravitational force and ground reaction force (GRF) – is equal to the sum of the mass-acceleration products of the body segments (Winter, 2009), the total GRF acting on a unilateral above-knee amputee with normative kinematics was calculated as

$$\text{GRF}_{\text{amp}} = \text{GRF}_{\text{able}}$$

$$\begin{bmatrix} m_d & m_{d_{\text{tot}}} & (\hat{\mathbf{p}}_{\text{d_{\text{tot}}}^\text{com}} + g\hat{y}) + \\ m_i & m_{l_{\text{tot}}} & (\hat{\mathbf{p}}_{\text{l_{\text{tot}}}^\text{com}} + g\hat{y}) + \\ m_f & m_{f_{\text{tot}}} & (\hat{\mathbf{p}}_{\text{f_{\text{tot}}}^\text{com}} + g\hat{y}) \end{bmatrix}$$

where $\text{GRF}_{\text{amp}}$ is the total GRF on the amputee; $\text{GRF}_{\text{able}}$ is the normative total GRF on an able-bodied human (Winter, 1991); $m_d$, $m_i$, and $m_f$ are the masses of the able-bodied upper leg, lower leg, and foot (Winter, 2009); $\hat{\mathbf{p}}_{\text{d_{\text{tot}}}^\text{com}}$, $\hat{\mathbf{p}}_{\text{l_{\text{tot}}}^\text{com}}$, and $\hat{\mathbf{p}}_{\text{f_{\text{tot}}}^\text{com}}$ are the COM accelerations of the upper leg, lower leg, and foot; $g$ is the gravitational constant; and $\hat{y}$ is a unit vector in the positive vertical direction. The GRF was prescribed to the model, and GRF during double-support was approximated based on a linear force transition from one leg to the other. Knee moment was then calculated, which represented the knee moment required for the amputee to walk with normative kinematics ($M_{\text{eq}}$).

Knee power was computed, and the gait cycle was observed to consist of 3 phases: 1 energy storage phase, and 2 energy dissipation phases (Figure 1).

![Figure 1: Phases of the gait cycle. $P_{\text{k}}^*$ is knee power normalized to body mass. Phase 1 is an energy storage phase, whereas phases 2 and 3 are energy dissipation phases.](image)

Based on this observation, a general mechanical model was defined consisting of a spring during phase 1 and dampers during phases 2 and 3. A cost function was defined during each phase equal to the least-squares error between the moment produced by the model ($M_{\text{mod}}$) and $M_{\text{eq}}$, and stiffness and damping coefficients were optimized to minimize the cost function.

RESULTS
An optimized first-order spring during phase 1 and optimized zero-order dampers during phases 2 and 3 were found to reproduce $M_{\text{eq}}$ with high accuracy (Figure 2).
EFFECTS OF PROSTHESIS MASS ON OPTIMAL STIFFNESS AND DAMPING PARAMETERS OF PROSTHETIC KNEES

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Figure 2: Comparison of $M_{req}$ and $M_{mod}$ over the gait cycle ($R^2 = 0.90$). $M^*$ is knee moment normalized to body mass, $M_{req}^*$ is $M_{req}$ normalized to body mass, and $M_{mod}^*$ is $M_{mod}$ normalized to body mass.

The stiffness of the first-order spring was found to be insensitive to changes in prosthesis segmental mass. The damping coefficients during phases 2 and 3 were highly sensitive, changing by up to 36% and 330%, respectively, as foot and lower leg mass varied between 25% and 100% of their corresponding able-bodied values (Figure 3). The damping coefficient during phase 2 was most sensitive to lower leg mass, whereas the damping coefficient during phase 3 was most sensitive to foot mass.

Figure 3: Effects of lower leg (shank) and foot mass on optimal coefficients. Top: stiffness coefficient during phase 1. Middle: damping coefficient during phase 2. Bottom: damping coefficient during phase 3. $m_f^*$ and $m_{ll}^*$ are foot and lower leg mass normalized to corresponding able-bodied values. Results are shown for upper leg mass equal to 25% of able-bodied value.

DISCUSSION

The results indicate that a simple mechanical model consisting of a first-order spring and 2 zero-order dampers can be used to accurately reproduce the moment required for an amputee to walk with normative kinematics. In a prosthetic knee, this model can be implemented using a linear spring and 2
friction dampers. In addition, damping coefficients were found to vary significantly with prosthetic mass. Thus, it is suggested that prosthesis components should not be optimized based on the moment-time relationship of an able-bodied human walking with normative kinematics, but that of an amputee.

CLINICAL APPLICATIONS
The results illustrate the parametric effects of mass on optimal stiffness and damping coefficients for prosthetic knees. These coefficients can enable a clinician or prosthettist to explain kinematic deviations of above-knee amputees, as well facilitate the design of passive prosthetic knees that utilize low-cost, mechanical components.

REFERENCES
Winter, D.A. Biomechanics and Motor Control of Human Movement. 2009.