Development of a Low-technology Prosthetic Swing-phase Mechanism

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Abstract

Although a number of accessible low-technology prosthetic devices have been developed, most only provide very basic function during the swing-phase. A prosthetic swing-phase control mechanism simulates the action of the upper leg musculature to aid in increased gait function. More specifically, swing-phase control mechanisms limit the maximum knee flexion and allow the shank to smoothly decelerate into full knee extension without excessive impact. In this study, a single above-knee amputee was fitted with a prosthetic knee, and different low-technology swing-phase setups were clinically assessed. Clinical testing included walk tests utilizing a mobile computer setup connected to electrogoniometers (to measure knee flexion angle and time) and accelerometers (to measure terminal impact decelerations) mounted on the sound and prosthetic limbs. As hypothesized, incorporating friction and a spring system improved gait function. The dual spring system, two springs in series, as predicted by our computational model and mechanical testing, out-performed the single spring system. The swing-phase knee torque versus flexion pattern of the dual spring system best matched baseline data from a high-end hydraulic swing-phase controller. Clinical trials revealed increased walking velocity, decreased and more normal maximum prosthetic knee flexion and lower terminal impact with the dual spring systems. The new dual spring mechanism is simple, improves the performance of a prosthesis, and is ideal for use in applications where size, weight and cost may be constraining factors. This includes the provision of prostheses to children and to individuals in developing countries.

Keywords: Prosthetics, Above-knee amputee, Swing-phase control, Gait, Biomechanics

1. Introduction

Developing countries have very high rates of amputation for many reasons; poor health care, sub-standard working conditions and unsafe methods of transportation can all lead to significant injury resulting in the loss of a limb [1]. Current and past zones of conflict exacerbate the issue since injuries from combat and residual land mines further increase the number of amputations. Disabilities are often amplified in developing countries, where health care and infrastructure may not be sufficient to accommodate those with disabilities. In developing countries, an estimated three to four million people require prostheses [1]. An affordable, highly-functional prosthetic knee joint, combined with an adequate distribution network and clinical/technical support could help many of those in need.

A prosthetic swing-phase mechanism simulates the action of the upper leg musculature to aid in improved gait function. More specifically, swing-phase mechanisms limit the maximum knee flexion and allow the shank to smoothly decelerate into full extension without excessive impact [2]. Without swing-phase control, numerous gait deviations can result, increasing energy demands and gait asymmetry [3].

Various systems have been developed to mimic the action of muscles that act about the knee joint. Swing-phase control mechanisms typically consist of friction brakes or dampers, extension bias or assist mechanisms and extension cushions. An extension assist mechanism typically takes the form of a mechanical spring, while damping is produced by a pneumatic or hydraulic cylinder. These devices aim to generate moments about the knee joint to allow the shank-foot to swing through space with a motion pattern which approximates that of an average able-bodied person [2].

Although pneumatic and hydraulic swing-phase control units approximate normal gait closely, they are larger and bulkier, which is a concern when fitting children [4]. They are also more expensive and require ongoing maintenance; this is an issue for patients in developing countries, who often live in

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rural areas and have to travel great distances to reach prosthetic repair centers [5]. Furthermore, failure of hydraulic knees can be difficult to assess by the patient, and early detection is challenging. Failures in hydraulic knee units can result in oil leakage and loss of function, bringing about embarrassing and dangerous situations for the amputee [6].

A number of accessible low-technology prosthetic devices have been developed, although most only provide basic function and lack technology that adequately assists the patient during swing-phase [7,8]. For example, the most widely used prosthetic system, the manually locking ICRC knee, permits only stiff legged gait, while the M1Knee and Jaipur knee are polycentric prosthetic limbs without extension assist systems. While other technologies such as the POF knee do incorporate some level of resistance through springs and/or friction, there are limited criteria used to ensure that the friction levels or spring stiffness are optimally set to provide adequate swing-phase control. For this reason, even existing low-cost swing-phase mechanisms (which typically utilize friction and springs) are severely limited in performance.

Numerous studies have been published focusing on the assessment of swing-phase control mechanisms. Most of these studies focus on the development of microprocessor-controlled knee units and compare their performance to conventional hydraulic and pneumatic systems [9,10]. The primary objective of this study was to show that low-technology swing-phase mechanisms (i.e., non-fluid-based systems) can be optimized to help improve gait function by allowing patients to achieve faster gait, lower heel-rise and more normal knee flexion, and decreased terminal impact. Furthermore, the goal was to gain quantitative data about the gait deviation associated with terminal impact. Several publications recognize the negative effects of terminal impact on amputees [11-13], although only a small number of studies have attempted to evaluate it [14,15], and only one study used quantitative methods [16].

2. Methodology

2.1 The design

The swing-phase control mechanism testing was performed using a prototype prosthetic knee joint (LC Knee) that is being developed at the Bloorview Research Institute. The LC Knee is based on a single-axis design and incorporates a novel stance-phase control mechanism to provide higher level function while maintaining stability during weight bearing [17]. As shown in Fig. 1, the LC Knee utilizes three mechanisms to achieve swing-phase control, including (1) friction to provide braking, (2) springs to provide extension bias, and (3) an elastomer to limit terminal impact.

Friction control is provided by tightening the nut on the bolt that passes through the center of the knee axis; this affects compression between the knee's articulating faces. The spring system also acts at the knee axis and produces an internal extension moment which helps in reducing maximum flexion and assists in knee extension. The LC Knee can be setup with one of two spring systems. One consists of a single

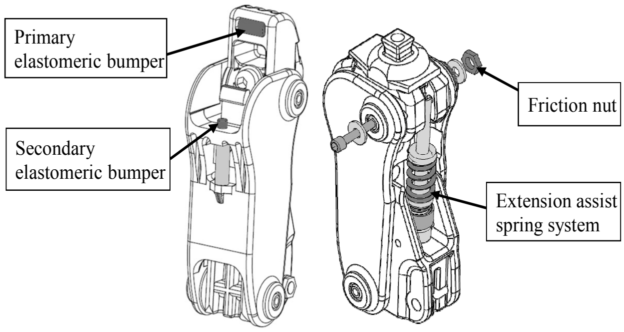

Figure 1. Features of the LC Knee swing-phase control mechanism.

compression spring, and the other is composed of a novel dual spring system. The dual spring system utilizes two springs in series with different spring constants to better approximate the action of the leg musculature during flexion and extension. Finally, the elastomeric bumpers reduce impact between the knee's contact surfaces at the end of swing-phase.

2.2 Computational model and mechanical testing

A computation model, coded in Matlab and developed at the Bloorview Research Institute, was used to optimize the mechanical swing-phase control of the LC Knee by matching it as best as possible to previously attained prosthetic hydraulic knee moment data [18]. These data were obtained using a kinematic simulator that mechanically simulated the action of swing-phase based on gait lab data obtained from an above-knee amputee at two walking speeds. The kinematic simulator setup can be seen in Fig. 2. The simulator outputs, torque and position, were used to plot knee torque versus flexion graphs. These data were then used in a computational model which predicted spring constants and friction levels for both the single and dual spring systems.

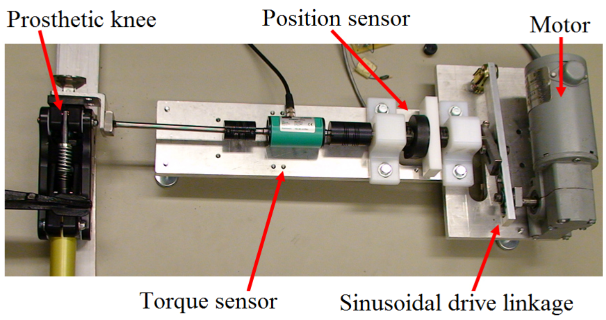

Figure 2. Kinematics simulator.

Within the computational model, the total knee torque of the system was calculated using the following formula:

$$\tau = (k_1 \ x \ \Delta x_1 \ x \ r) + (k_2 \ x \ \Delta x_2 \ x \ r) + (F_{pc} \ x \ r) + \tau_f \qquad (1)$$

where r is the moment arm the spring acts upon about the center of the knee axis, k_1 and k_2 are the spring constants of each spring, Δx_1 and Δx_2 are the amounts of deflection of each spring, F_{pc} is the pre-compression force in the spring system, and τ_f is the torque due to friction.

In order to solve for the aforementioned variables, linear optimization was performed to minimize the root mean square

(RMS) error between the hydraulic and friction/spring mechanism torque data. This was achieved using an incremental and sequential assignment of values to the aforementioned variables from allowed sets. In this way the outputs (total knee torque produced by the friction/spring mechanism) for all possible combinations of variable values were assessed and compared to the hydraulic data. The allowed sets of variables included the range of possible (and technically feasible) friction levels (0 and up), spring constants, and precompression values. For example, the spring constant variable was limited to a range of 0 to 100,000 N/m, since a stiffer spring (>100,000 N/m) with the necessary deflection would be unpractical for use in a prosthesis.

A friction control knob for the LC Knee was created to allow more precise control of the friction levels during clinical testing. By tightening the knob to set incremental values and examining the subsequent changes in torque output from the kinematics simulator, we were able to reliably control and set friction levels.

2.3 Clinical testing

To validate the results of the computational model, clinical testing with an above-knee amputee was completed. The participant was 18 years old, weighed 72.5 kg and was 177 cm tall. The participant was fitted with a LC Knee and allowed to use the prosthetic leg for a month prior to testing. The clinical testing included a series of 20-m walk tests utilizing a mobile computer setup connected to electrogoniometers (Biometrics Ltd. SG150) and an accelerometer (Silicon Designs Inc. 25G) mounted on the sound and prosthetic limbs. The goniometers measured knee flexion angle, and the accelerometer measured terminal impact accelerations. Walking speed was measured using a stop watch. The participant completed twenty trials with the different swing-phase setups at two walking speeds, his regular, self-selected walking speed, and his fast walking speed. The swing-phase setups included two friction levels, incorporating a secondary extension bumper and incorporating five different spring systems: no spring (NS), single spring (SS1), single stiff spring (SS2), dual spring (DS1) and dual stiff spring (DS2). It is important to note here that, for the NS condition, a high friction condition was not tested since this condition was deemed unsafe for the participant due to the difficulty of fully extending the leg in late swing-phase. In addition, the low friction/no spring (NS) condition is a relevant baseline to compare to since it is most representative of existing technologies (M1Knee, Jaipur knee, ICRC knee etc.) that do not incorporate mechanisms to provide swing phase control and act primarily as a free hinge. Ethics for the study were approved by the Bloorview Research Ethics Board, and the participant provided written informed consent.

3. Results

3.1 Computational model and mechanical testing

The program recommended spring constants and friction levels at the two walking speeds for both spring systems (single and dual). Table 1 displays a summary of the optimized output. Figure 3 illustrates the torque-versus-flexion curves of the optimized single and dual spring systems, hydraulic prosthetic knee and friction only setup at self-selected walking speed. The program results confirmed the hypothesis that a dual spring system could better match the torque curve of a hydraulic system when the root mean square difference was minimized. DS1 with the 1500 N/m and 19,500 N/m springs performed best when considering both self-selected and fast walk walking speeds. Using the results of the computational model and taking into consideration past experience with the LC knee, four spring systems were selected and two levels of friction, low (0.75 Nm) and high (1.5 Nm), were established for clinical testing (see Table 2).

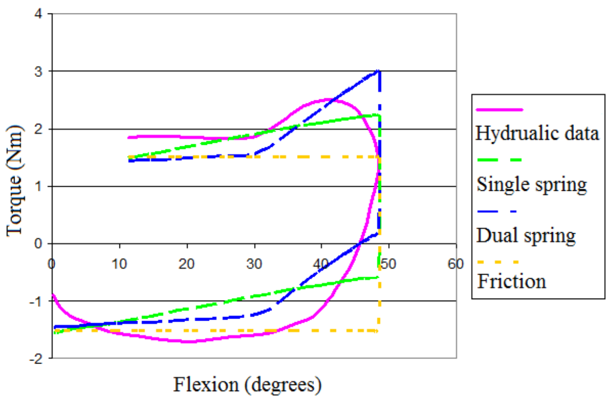

Figure 3. Program output – optimized spring systems for swing-phase knee flexion.

Table 1. Computation model output summary.

Spring system	Walking speed	Spring constant (N/m)	Friction torque (Nm)
Single	Self-selected	5000	1.4
	Fast walk	7500	1.8
Dual	Self-selected	1500 & 19500	1.4
	Fast walk	3500 & 19500	1.8

Table 2. Experimental system values.

	Spring constant (N/m)	Friction torque (Nm)
NS	-	0.75
SS1	4205	0.75 / 1.5
SS2	8585	0.75 / 1.5
DS1	1051 & 19447	0.75 / 1.5
DS2	1051 & 41689	0.75 / 1.5

3.2 Clinical Testing

The results indicate that swing-phase mechanisms greatly aid in improving gait. Adding extension assist springs and friction helps increase walking velocity, decrease maximum knee flexion and reduce terminal impact. Figure 4 compares the maximum flexion and terminal impact of the spring systems and demonstrates the effects of increasing friction. Table 3 displays the percent improvement of incorporating swing-phase control for self-selected and fast walking speeds.

Condition DS2 with high friction performed best overall, with the participant attaining a 16.8% (0.19 m/s) increase in

Table 3. Clinical testing summary.

Swing-phase control	Walking speed	Velocity (m/s)		Maximum flexion (degrees)		Terminal impact (g)	
		Mean	PI	Mean (SD)	PI	Mean (SD)	PI
NS	SS	1.13	-	84.2 (±2.1)	-	26.8 (±2.2)	-
	FW	1.53	-	90.6 (±1.0)	-	26.0 (±3.0)	-
SS1	SS	1.31	15.9%	69.7 (±1.4)	17.2%	19.2 (±1.4)	28.3%
	FS	1.59	4.2%	77.1 (±1.3)	14.9%	24.1 (±1.0)	7.1%
SS2	SS	1.45	28.3%	66.2 (±1.5)	21.4%	18.1 (±1.0)	32.5%
	FW	1.64	7.2%	74.2 (±2.0)	18.1%	18.7 (±0.8)	28.1%
DS1	SS	1.35	19.6%	68.2 (±1.3)	19.0%	17.1 (±0.8)	36.2%
	FW	1.63	6.5%	76.4 (±1.6)	15.7%	18.9 (±1.8)	27.5%
DS2	SS	1.32	16.8%	63.9 (±0.9)	24.1%	13.9 (±0.7)	48.1%
	FW	1.72	12.4%	74.0 (±0.8)	18.3%	17.9 (±2.0)	31.2%

Note: No spring condition (NS) – 0.75 Nm friction; spring conditions (SS#/DS#) – 1.5 Nm friction.
SS – self-selected; FW – fast walk; PI – percent improvement vs. NS

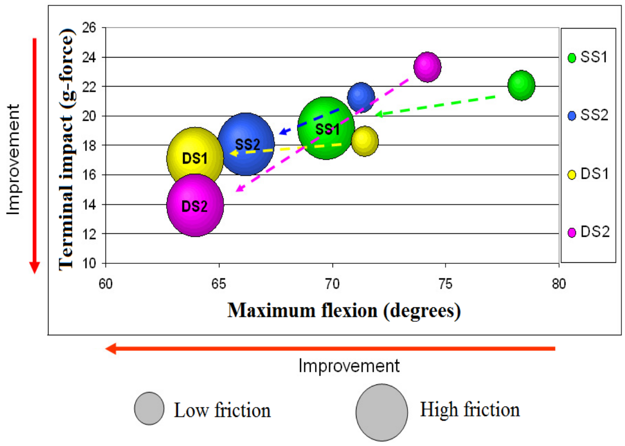

Figure 4. Spring system evaluation – friction.

velocity. Maximum flexion decreased on average by 24.1% (20.3 degrees) to provide more normal peak knee flexion angle of 63.9 degrees [19]. Furthermore, this reduction in prosthetic heel-rise helped to better match the intact limb kinematics, significantly improving gait symmetry. Condition DS2 with high friction reduced terminal impact by 48.1% (12.9 g) compared to the NS condition and by 27.6% (5.3 g) compared to the SS1 condition (see Fig. 4). Removing the secondary elastomeric bumper resulted in 13.9% (3.4 g) higher terminal impact decelerations.

4. Discussion

The computational model recommended that the less stiff dual spring (DS1) provided optimal control, while the stiffer dual spring (DS2) performed best during clinical testing. This discrepancy is due to the fact that participant adopted faster walking speeds with the LC Knee than when using the hydraulic knee. Using the hydraulic knee (on which the computational model is based) the participant attained a self-selected walking speed of 1.05 m/s and a fast-walk walking speed of 1.43 m/s. With the LC Knee, the user was able to attain a self-selected walking speed of 1.32 m/s and a fast-walk walking speed of 1.72 m/s. The increase in walking velocity (with the LC Knee) resulted in the need for higher knee torques [18], and therefore a stiffer spring.

Increased walking velocity is a common goal of many areas of rehabilitation, as it is considered indicative of overall improvements in mobility function [20,21]. Increased gait velocity is also associated with higher-end prosthetic components [22,23]. In this study, when adjusted to closely match the torque profile of a high-end fluid-based system, the low-technology swing-phase control mechanism increased walking speed, produced more symmetrical gait, decreased unwanted heel-rise, and reduced terminal impact.

The dual spring system, two springs in series, as predicted by a computational model outperformed the single spring system. The dual spring system's greatest improvement was in lowering terminal impact. Reduced terminal impact indicates smoother swing and smoother transition from swing-phase to stance-phase [16]. This is achieved by the deactivation of the stiff spring and activation of the less stiff spring during the last 20 degrees of swing-phase before full knee extension to allow the shank to decelerate and impact the bumper at a lower velocity. The activation/deactivation of the dual spring is possible due to a hollow washer stop that surrounds the less stiff spring, which forces the spring to prematurely bottom out at a predetermined displacement. Further examination of the angular velocities derived from the potentiometer data verifies this hypothesis (see Fig. 5). The foot-shank averaged 520 degrees/s with the single spring systems and 484 degrees/s with the dual spring system, a reduction in angular velocity of 7%. Providing sufficient shock absorption to amputees through the addition of specialized prosthetic components, such as shock absorbing pylons, has been shown to increase comfort and gait performance and prevent joint and back problems in the long term [24]. Therefore, the same benefits may potentially be realized by reducing terminal impact.

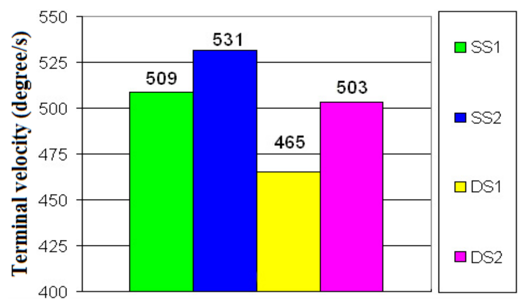

Figure 5. Average terminal swing velocity.

It is evident from the study results that friction also plays an important role in enhancing gait characteristics. The high friction condition resulted in both lower (and more normal) maximum flexion and terminal impact while enabling the user to maintain high walking velocities. These results agree with previously published studies [25,26], but also suggest that friction on its own is not adequate in providing swing-phase control. Effective non-fluid-based swing-phase control requires the combination of friction and spring systems, which are adjusted to match the physical characteristics of the prostheses, as well as functional requirements of the prosthetic user.

In this study, performance of the low-technology swing-phase control was enhanced by adjusting various parameters including the braking friction, spring constants, and spring precompression to provide a torque profile to match that of high-end fluid-based systems. In clinical practice, while it may not be feasible to apply this approach on a per-patient basis; future work should focus on developing and providing prosthetists with guidelines as how to systematically adjust parameters to achieve best gait performance. This involves clearly defining the relationships between patient characteristics (age, health, experience), swing-phase mechanism parameters (friction level, spring forces) and gait parameters (walking speed, heel-rise, terminal impact).

Although the prosthetic limb was tuned to a single individual, as seen in Table 3, the consequences of improper tuning (or lack of swing-phase control) can substantially reduce walking speed and introduce gait deviations (i.e. excessive heel rise and terminal impact). Moreover, mitigation of gait deviations has been associated with lower energy consumption [19]. Future work should involve more subjects to ensure a better representation of the population and confirm these preliminary results. Clinical testing should also evaluate conventional and other low-cost prosthetic knee joints, and include physiological measures to assess the effects of swing-phase control on energy consumption.

5. Conclusions

As hypothesized, adjusting a friction and spring system to more closely match the torque characteristics of a high-end fluid-based damper resulted in improved prosthetic gait function. The new dual spring mechanism is simple, improves prosthetic function, and is ideal for use in prostheses where size and cost may be constrained. Future work will aim to apply a larger sample size to investigate the generalizability of these results.

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