Improving the Gait Performance of Nonfluid-Based Swing-Phase Control Mechanisms in Transfemoral Prostheses

Alex Furse, William Cleghorn, and Jan Andrysek*

***Abstract*—A prosthetic swing-phase control mechanism simulates the action of leg musculature, aiding gait function by controlling the duration of swing, extent of heel rise, and by allowing the shank to smoothly decelerate into full knee extension without excessive impact. Nonfluid-based (NBF) mechanisms have the potential to provide a durable and affordable solution as required in many parts of the world, but the design variables that lead to improved performance of NBF swing-phase control technologies are not well established. Seven transfemoral amputees were fitted with a prosthetic knee joint and different NBF swing-phase setups were systematically assessed. Clinical testing included walk tests utilizing a potentiometer (to measure knee flexion angles) and accelerometer (to measure terminal impact decelerations) mounted on the prosthetic limb. As hypothesized, the friction and spring systems improved gait function. This includes an increased walking speed that closely matched high-end hydraulic prosthetic knee joints, decreased and more normal maximum prosthetic knee flexion, decreased flexion duration, and lower terminal impact. Further improvements were obtained using a dual-spring system, two springs in series, over the more conventional single spring system. NBF swing-phase control mechanisms are simple and significantly improve the performance of prostheses. Their application is ideal where size, weight, and cost may be constrained.**

***Index Terms*—Above knee, amputation, amputee, control, gait, knee joint, prostheses, prosthetic, swing, swing-phase control, transfemoral.**

I. Introduction

SIGNIFICANT progress has been made in the development of technologically advanced prosthetic devices that can enable individuals with lower-limb amputations to achieve highly functional mobility. In terms of prosthetic knee joint technologies, these include sophisticated pneumatic or hydraulic dampers that can be controlled using electronics. These dampers serve an important function during the swing phase of gait, and generate passive moments or torques about the knee joint to cause the shank-foot complex to swing through space with a kinematic pattern, which approximates that of an average able-bodied person [1], [2]. In particular, the swing-phase control mechanism works to: 1) limit the maximum knee flexion; 2) shorten the duration of swing; and 3) smoothly decelerate the shank into full extension without excessive impact [3]–[8]. For active individuals, it is important that these functions are adequately performed over a range of walking speeds. Without effective swing-phase control, numerous gait deviations can result increasing energy demands and gait asymmetry [9]–[11].

The simplest method of producing a passive torque at the knee joint is by using a friction braking mechanism. Whereas friction can help to reduce excessive knee flexion, the drawback is that the knee joint can only be properly adjusted for one slow walking speed [3], [12]. Friction-based swing-phase control is, therefore, prescribed primarily for less active individuals. Occasionally, elastic members such as springs are used to assist the extension of the knee [4], [7]. Hence, this type of mechanism is commonly termed an extension bias or extension assist. Depending on the specific setup, these mechanisms can aid in decreasing the swing time and maximum knee flexion and increasing cadence by forcing the knee into extension. But this has limits and providing more extension assist can result in excessive terminal impact [7]. To a degree, terminal impact can be managed by using compliant extension cushions or bumpers. These relationships between gait variables and variables associated with nonfluid-based (NFB) swing-phase mechanisms are summarized in Table I.

More sophisticated and commonly used swing-phase control mechanisms are fluid based, including hydraulic and pneumatic dampers [4], [12], [13]. In contrast to NFB systems, fluid-based systems are cadence responsive and capable of increasing damping resistance at faster walking speeds to provide better control over a range of walking speeds (see Fig. 1) [14]. However, whereas fluid-based technologies provide superior swing-phase control and are clearly indicated for active individuals, when compared to NFB systems they are substantially more expensive and, therefore, are typically inaccessible for active individuals residing in the many under-resourced parts of the world. They are also costly to repair and maintain [4], [12]. This is a real problem for individuals who live in rural areas and have to travel great distances to reach prosthetic repair centers [15], [16].

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A. Furse was with the Institute of Biomaterials and Biomedical Engineering, University of Toronto, ON M5S 3G8, Canada. He is now with the Princess Margaret Hospital, University Health Network, Toronto, ON M5G 2C4, Canada (e-mail: alex.furse@utoronto.ca).

W. Cleghorn is with the Department of Mechanical and Industrial Engineering, University of Toronto, ON M5S 3G8, Canada (e-mail: cleghrn@mie.utoronto.ca).

*J. Andrysek is with the Institute of Biomaterials and Biomedical Engineering, University of Toronto, ON M5S 3G8, Canada, and also with the Bloorview Research Institute of Holland Bloorview Kids Rehabilitation Hospital, Toronto, ON M4G 1R8, Canada (e-mail: jan.andrysek@utoronto.ca).

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TABLE I
RELATIONSHIPS BETWEEN SWING-PHASE MECHANISM VARIABLES AND GAIT VARIABLES

	Add Friction	Add spring for extension bias	Add extension bumper
Increase walking cadence/speed	o	+ [7]	o
Decrease heel-rise/peak knee flexion	+ [6]	+ [4]	o
Decrease duration of swing-phase	o [6]	+ [5;7]	o
Decrease terminal impact	+	- [4;7;8]	+ [4]

+ signifies a positive effect, - signifies a negative effect, o signifies no effect or unknown effect. Reference numbers shown in square brackets.

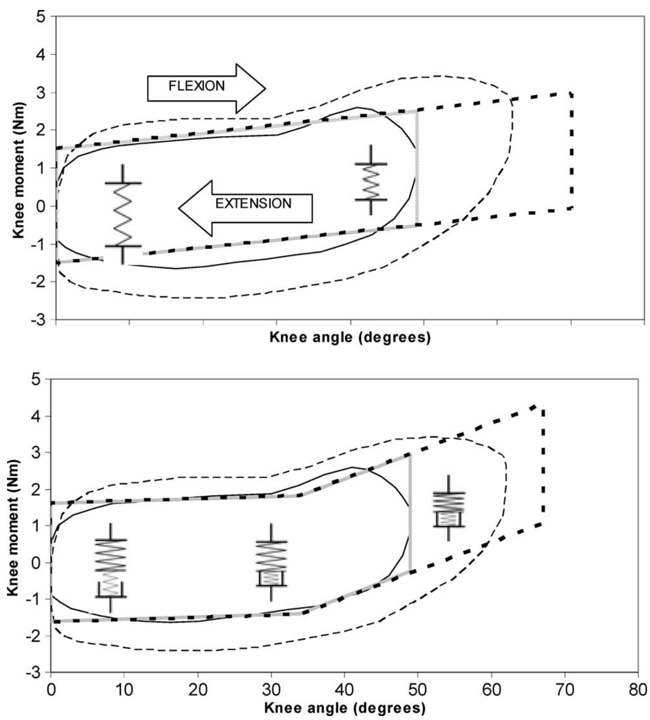

Fig. 1. Flexion angle versus knee moments for a hydraulic swing-phase damper for self-selected walking speed (——) and fast walking speed (- - - - - -). Single spring system with friction shown in top graph and dual-spring system with friction in bottom graph. The dual-spring system which increases stiffness at a greater rate during high knee flexion angles provides more torque at faster walking speeds to better limit excessive heel rise and flexion duration. Self-selected walking speed for the NFB system shown as grey thick line (——), and fast walking speed as thick dashed line (– – –). Note that the dual-spring system better matches the torque profile of the hydraulic system, and is also capable of providing higher torques at higher knee flexion angles. The data in this figure assume an ideal mechanism that directly applies the spring force to produce knee torques and is based on previous work [21].

Fluid-based systems are considerably more complex than NFB systems, resulting in lower durability; this along with increased bulkiness and weight generally makes them unsuitable for use in pediatric prostheses [17]. Hence, conventional practice in under-resourced regions and in pediatrics alike is not to use fluid-based systems in prosthetic knee joints [18]–[20]. Whereas occasionally swing-phase control is omitted altogether, in most cases swing-phase control is provided, either as a friction mechanism, an extension assist mechanism, or both [18]–[20]. Hence, the application of NFB swing-phase control mechanisms is not only largely varied in clinical practice, but these mechanisms are also often either underutilized, or just not effectively utilized. At least part of this can be attributed to the limited number of studies that have systematically investigated the functional characteristics of NFB mechanisms to allow for their optimal utilization in clinical practice [3], [6].

As part of the overarching goal of this study, which is to establish a scientific basis for the design and effective utilization of NFB swing-phase control mechanisms in lower-limb prosthetics, the specific focus of this study was to systematically explore the effects of common design variables associated with NFB swing-phase control technologies (friction, extension bias or assist, and extension cushions) on core prosthetic swing-phase gait variables including walking speed, swing-phase kinematics, and terminal impact decelerations. We hypothesized that a NFB swing-phase mechanism will improve gait variables as outlined in Table I.

Furthermore, earlier work suggested that swing-phase control can be enhanced by using a dual-spring system, two springs in series, whereby a less stiff spring provides lower torques during small knee flexion angles and a stiff spring provides higher torques at higher knee flexion angles, to better match the characteristics of a fluid-based damper at slower and faster walking speeds, respectively (see Fig. 1) [21]. It was, therefore, an additional goal of this study to clinically assess the effectiveness of the dual-spring system as compared to more conventional single spring systems when used by active individuals. Hence, our second hypothesis was that a dual-spring mechanism will better control heel rise, swing-phase duration, and decrease terminal impact, when compared to a conventional single spring system.

II. METHODOLOGY

A. Participants and Prostheses

A convenience sample of seven active (K3 and K4 level) individuals (6 males and 1 female) with unilateral transfemoral ($n = 6$) and knee disarticulation ($n = 1$) amputations were selected for this study. Average age of the participants was 36.5 years (range 18–58 years). Average height of the participants was 178 cm (range 171–184 cm) and average weight was 84.0 kg (range 60–106 kg).

The conventional prostheses used by the participants were composed of the following components. Prosthetic feet used were the Axion (Otto Bock), Variflex and Elation (Össur), Multiflex (Endolite), and Seattle Carbon Lighfoot (Kinsley). Knee joints included the Total Knee 2000, Total knee 2100 and Mauch SNS (Össur), EUK and KX06 hydraulic knees (Endolite), and 3R60 (Otto Bock). Four of the participants used suction suspension, two used waist belts, and the individual with the knee disarticulation used self-suspending supracondyler buildups. For testing of the NFB swing-phase control systems, a duplicate prosthesis was made for all but one participant for whom testing was done using his conventional prosthesis by exchanging knee joints. For these testing prostheses, sockets were duplicated and components were matched to those of the conventional

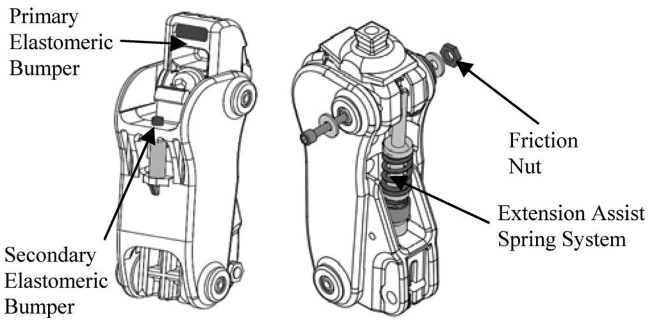

Fig. 2. SASPL knee and components of the swing-phase control system showing frontal-side view (left) and posterior-side view (right).

prostheses, with the exception of the prosthetic feet; Seattle feet (Kingsley) were used in all cases. Testing of the NFB swing-phase control systems was done using a stance-phase-controlled prototype knee joint, termed the simplified automatic stance-phase lock (SASPL) knee [22]. All setups and alignments were performed by a certified prosthetist. The participants were fitted with the testing prosthesis and given at least 2 weeks of acclimation prior to data collection. Informed consent was obtained from the participants prior to participation in the study, as approved by the research ethics board of the facility.

B. Testing Protocol

The study consisted of a repeated measures design to evaluate different NFB prosthetic swing-phase control conditions. As described in detail later, these were applied using three mechanisms including a friction mechanism, an extension assist mechanism, and an extension cushion (see Fig. 2). The use of the SASPL knee was important as it provided a standardized method for applying these testing conditions.

For the friction mechanism, tightening the nut on the bolt that passes through the center of the knee axis of the SASPL knee enabled adjustment of the compression between the articulating surfaces, thereby, allowing the friction level to be increased. Two friction levels were evaluated. To allow for the accurate adjustment of the low and high friction levels, a friction control knob was created and used during the clinical testing. A high friction level of 1.5 Nm and a low friction setting of 0.75 Nm were established and verified for consistency by measuring the friction torque of three different SASPL knee joint prototypes. Consistency was found to be acceptable and within 0.15 Nm of the desired values.

The magnitudes of the friction levels were based on clinical experience. Particularly, the higher friction level of 1.5 Nm was determined to be well tolerated by the participants of this study in prior clinical testing. However, levels above this were found to be unacceptable for some of the participants, as they felt that there was too much resistance and that they could not adequately extend the leg during swing phase to ensure safe weight acceptance. A low friction level of 0.75 Nm was arbitrarily selected at 50% of the high level. During clinical testing the friction levels were applied by tightening the friction control knob a predetermined amount.

For the extension assist mechanism, the SASPL knee was setup with one of two spring systems. The first consisted of a single compression spring, and the second was composed of a dual-spring system. The dual-spring system utilizes two compression springs in series with different spring constants to better approximate the action of the leg musculature during flexion and extension [21]. In effect, a less stiff spring is active from full knee extension to approximately 30° of knee flexion, and after that a stiffer spring becomes active (see Fig. 1). Low and high stiffness conditions for each spring system (single and dual) were evaluated. These were based on earlier work and the clinical experience with the SASPL knee [21].

Extension cushions consisting of elastomeric bumpers were used to reduce impact between the knee's contact surfaces at the end of swing phase. The primary bumper provided a relatively stiff stopping surface, which is needed to provide a definitive knee extension limit. This bumper was applied during all testing conditions. The secondary bumper, which was substantially more compliant, was designed to come into effect shortly ($\approx 4°$) before full knee extension, thus providing a more gradual deceleration. This bumper was used under all testing conditions except for those designated as having "no bumper," for which it was removed.

The clinical testing included a series of 20-m walk tests. Within a single session the participant completed 20 trials with the different swing-phase conditions at two walking speeds, including self-selected and fast walking. The swing-phase control setups included changing friction levels (High and Low), incorporating five different spring systems (No Spring, Single Spring, Single Stiff Spring, Dual Spring, and Dual Stiff Spring), and incorporating a secondary extension bumper (Bumper and No Bumper). Spring constants used were: Single Spring—4205 N/m, Single Stiff Spring—8585 N/m, Dual Spring—1051 and 19447 N/m, and Dual Stiff Spring—1051 and 41689 N/m. A breakdown of the experimental structure is found in Fig. 3. The order of testing was randomized. The walking tests were also performed with the participant's conventional high-end prosthesis.

C. Instrumentation

During the 20-m walk tests data were collected with a mobile computer setup connected to a potentiometer and an accelerometer mounted on the prosthetic limb (see Fig. 4). The data acquisition system consisted of a small laptop (ASUS EEEPC8G-W010) connected to a data acquisition board (National Instruments DAQ—USB-6212). The data were sampled at 2 kHz using a program written in Labview 7.1. Walking speeds were measured using a stop watch. A single axis 25 g accelerometer (Silicon Designs Inc., 2260) was used to sample the terminal impact decelerations (g-forces). It was mounted using adhesive tape 10 cm from the bottom of the prosthetic knee with its sensing axis oriented in the anterior direction of the shank segment of the limb. The potentiometer (Novotechnik—SP2801 308) was rigidly mounted to the rotating shaft. Knee angle data were used to calculate maximum knee flexion, terminal swing-phase velocity, and flexion duration. The aforementioned instrumentation

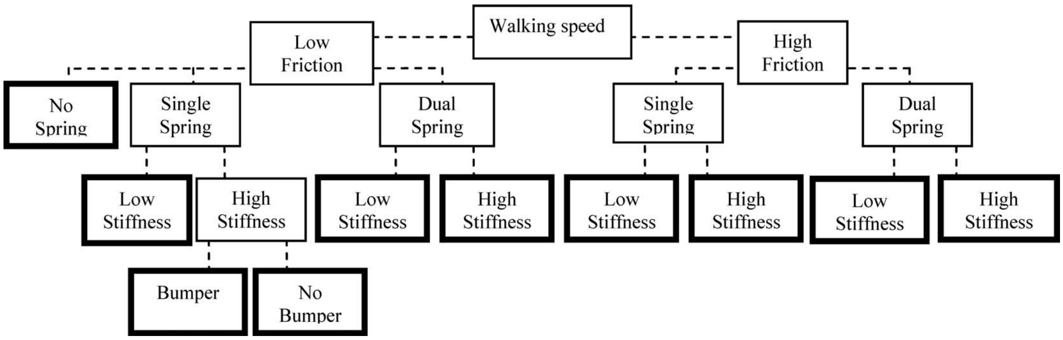

Fig. 3. Experimental structure breakdown. The above ten testing conditions were completed in randomized order for both self-selected and fast walking speeds.

Fig. 4. Testing prosthesis instrumented with potentiometer and accelerometer connected to laptop and data acquisition system

was not applied to the conventional prostheses. For the accelerometer the problem was due to the variety of prostheses and the presence of a compliant cosmetic cover in some cases (which attenuated the accelerations), it was not possible to obtain reliable signals. Similar issues were realized in trying to use the electrogoniometer. Therefore, only walking speeds were measured for the conventional prostheses.

D. Data Analysis

For each condition, the initial two cycles were left out of the analysis to ensure that steady-state pace was achieved and the subsequent five swing-phase cycles were analyzed. From the potentiometer data, variables that were extracted included: maximum flexion peaks, terminal swing-phase velocity which was determined by calculating the knee angle derivatives during the final 5° to 15° of swing phase before full knee extension, and flexion duration which was calculated by taking the temporal difference at the 5° flexion marks between early swing phase and late swing phase. These measurements were taken at 5° to avoid the noise present in some trials around 0° [23]. In addition, the peak decelerations at terminal impact, seen as a sharp spike corresponding to full knee extension at the end of swing phase, were extracted from the accelerometer data.

Hierarchical linear regression analysis using Statistical Analysis System software was used to control for within-subject correlation of repeated measures data. To evaluate the various swing-phase setups, the least-squares estimate difference (LSED) with 95% confidence intervals were calculated. The LSED is the estimate of the effect of changing the levels of one factor after controlling for all other factors in the model.

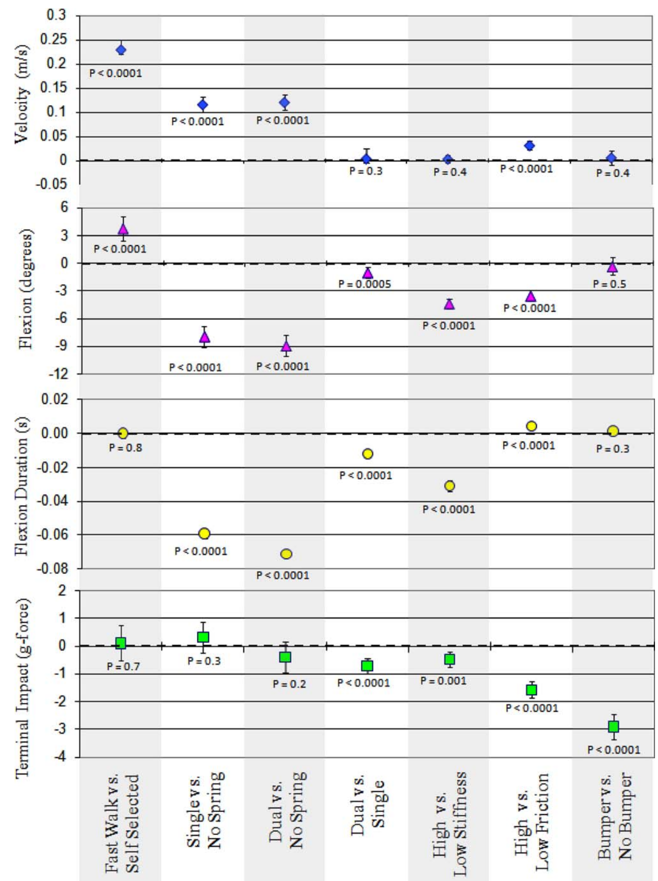

Fig. 5. Statistical analysis results. LSED with 95% confidence intervals ($n = 7$). The graph shows the effects of incorporating different swing-phase mechanisms on gait measures.

III. Results

The results of the hierarchical regression analysis are shown in Fig. 5. The figure displays the LSED of various swing-phase setups with 95% confidence intervals. The p-values of each comparison are also indicated. Table II displays the average walking speed, flexion duration, maximum flexion, and terminal swing-phase impact at each friction level for both walking speeds. One subject was unable to complete the fast walk trials and data for the high friction conditions of another subject were not usable because of instrumentation failure.

TABLE II
TESTING RESULTS SUMMARY

Swing-phase Control System Configuration	Walking Speed		Flexion Duration		Max Flexion		Terminal Impact	
	Mean (m/s)	± 1 SD	mean (s)	± 1 SD	Mean (degrees)	±1 SD	Mean (g force)	±1 SD
Conventional Fluid-based Knee Joints								
Self Selected Walking Speed	1.18	0.12						
Fast Walking Speed	1.49	0.22						
Low Friction – Self Selected Walking Speed								
No Spring	1.02	0.13	0.51	0.03	80.3	11.5	12.0	7.7
Single Spring	1.14	0.12	0.45	0.02	73.2	10.1	13.6	6.5
Single Stiff Spring	1.12	0.12	0.42	0.02	68.3	8.9	12.7	7.5
Single Stiff Spring (NB)	1.13	0.12	0.41	0.03	68.6	6.6	16.4	6.4
Dual Spring	1.14	0.10	0.44	0.02	72.0	8.8	12.2	6.4
Dual Stiff Spring	1.14	0.12	0.40	0.02	68.8	6.8	11.6	7.8
Low Friction – Fast Walking Speed								
No Spring	1.27	0.21	0.51	0.03	88.0	9.3	14.9	6.2
Single Spring	1.40	0.15	0.45	0.03	84.0	10.8	17.3	7.7
Single Stiff Spring	1.39	0.15	0.42	0.02	78.9	7.7	16.3	7.2
Single Stiff Spring (NB)	1.39	0.12	0.42	0.03	78.2	8.5	19.5	6.5
Dual Spring	1.38	0.13	0.43	0.02	81.2	10.2	15.7	7.2
Dual Stiff Spring	1.42	0.16	0.40	0.02	77.5	8.4	16.2	8.7
High Friction – Self Selected Walking Speed								
Single Spring	1.17	0.11	0.45	0.02	69.4	7.2	11.7	6.3
Single Stiff Spring	1.16	0.15	0.42	0.02	64.0	8.7	11.0	6.8
Dual Spring	1.21	0.16	0.45	0.02	67.9	8.9	11.0	6.1
Dual Stiff Spring	1.17	0.12	0.42	0.01	64.5	7.2	9.3	5.3
High Friction – Fast Walking Speed								
Single Spring	1.42	0.13	0.46	0.02	77.3	7.9	16.2	8.3
Single Stiff Spring	1.44	0.16	0.43	0.03	72.9	8.6	14.7	7.3
Dual Spring	1.43	0.16	0.45	0.03	76.4	9.0	14.4	6.6
Dual Stiff Spring	1.44	0.17	0.42	0.01	71.9	8.0	13.7	7.2

A. *Walking Speed*

Incorporating a swing-phase control mechanism was found to significantly increase the walking speed ($p < 0.0001$). On average the participants attained a 10.5% or 0.12 m/s increase in walking speed when a spring system was integrated (see Fig. 5). Compared to the conventional prostheses, walking speeds were 3.0% (self-selected) and 3.6% (fast walking) slower with the best performing NFB systems (i.e., dual stiff spring with high friction). In contrast, the No Spring condition produced 13.2% (self-selected) and 15.0% (fast walking) slower speeds when compared to the conventional prostheses. The difference between single and dual-spring systems was not significant ($p = 0.3$). Spring stiffness also had no effect on walking speed ($p = 0.4$). Friction, however, had an effect ($p < 0.0001$) allowing participants to attain 0.03 m/s faster gait speed with higher friction. As expected, removing the bumper had no effect on walking speed ($p = 0.4$). The fast walking speed was on average about 0.23 m/s faster than the self-selected walking speed (see Fig. 5). Three of the participants were unable to comfortably walk at a faster pace under the No Spring condition.

B. *Maximum Knee Flexion*

The amount of maximum swing-phase flexion across all conditions ranged from 64° to 88° (see Table II). Both friction and the spring systems significantly contributed to decreasing knee flexion ($p < 0.0001$) (see Fig. 5). The spring systems on average decreased the amount of flexion by 11.7% or 8.4° ($p < 0.0001$) (see Fig. 5). When compared to the single spring system, the dual spring performed significantly better to reduce maximum knee flexion ($p = 0.0005$). The stiffer spring systems performed significantly better than the low stiffness springs ($p < 0.0001$). Increasing the friction resulted in a mean decrease in knee flexion of 3.5° ($p < 0.0001$) (see Fig. 5). As expected, the extension bumper had no effect on maximum flexion ($p = 0.5$). Faster walking speed increased knee flexion by 3.8° ($p < 0.0001$) on average (see Fig. 5).

C. *Knee Flexion Duration*

Incorporating springs decreased the flexion duration by approximately 16.2% or 0.07s ($p < 0.0001$) (see Fig. 5). The dual-spring system achieved significantly lower values of flexion duration compared to the single spring system ($p < 0.0001$). The high stiffness springs outperformed the low stiffness springs ($p < 0.0001$). Increasing friction had a small but significant effect on increasing flexion duration ($p < 0.0001$). Presence of the secondary extension bumper had no effect on flexion duration ($p = 0.3$). Walking speed had no effect on flexion duration and the difference between the two walking speeds was insignificant ($p = 0.8$) (see Fig. 5).

D. Terminal Impact

Of the different swing-phase control configurations removing the secondary extension bumper had the greatest effect on terminal impact, resulting in decelerations that were 19% or 2.9 g higher ($p < 0.0001$) (see Fig. 5). Increasing friction lowered the terminal impact by 1.6 g on average ($p < 0.0001$). The dual-spring system produced lower decelerations than the single spring system ($p < 0.0001$) and the higher stiffness spring also significantly lowered decelerations ($p = 0.001$) although these effects were on average typically less than 1 g. Increasing walking speed had no effect on terminal impact decelerations.

IV. Discussion

The clinical utilization of simple NFB swing-phase control mechanisms is well documented in the literature [3]–[7]. This information, however, is in large part based on anecdotal clinical experience rather than systematic, empirically based studies. As such, the effectiveness of the various parts of NFB mechanisms toward improving gait function is not well understood. The findings of this study are important and may be utilized in the development of more effective NFB swing-phase mechanisms, as well as in clinical practice to optimize the performance of existing systems.

Increased walking speed is a common goal of many areas of rehabilitation. Furthermore, it is considered indicative of overall improvements in mobility function and is associated with the use of higher end prosthetic components [24]–[28]. In this study, the increased gait speeds attained after incorporating swing-phase control mechanisms were clearly evident. Compared to the no spring condition with low friction, incorporating a spring system with high friction caused the average self-selected walking speed to significantly increase from 1.02 to 1.14 m/s. Although this is still lower than typical able-bodied gait which on average is closer to 1.4 m/s [29], it is encouraging because the NFB system was only marginally (0.04 m/s) slower than the 1.18 m/s achieved with the high-end fluid-based knee joints in this study, and also other studies. Specifically, slower than normal mean self-selected walking speeds were found by Seroussi *et al*. of 1.20 m/s [30], Johansson *et al*. of 1.14 to 1.20 m/s [31], Blumentritt *et al.* of 1.30 m/s [32], and Segal *et al*. of 1.21 to 1.31 m/s [11] for high-end fluid-based knee joints. Moreover, a statistically significant difference of 0.10 m/s in self-selected walking speed between two types of fluid-based knee joint technologies has been recognized as being clinically significant [11]. This suggests that NFB swing-phase control is effective at enabling faster and more normal gait speeds and, therefore, these systems may play an important role in the rehabilitation of certain populations of active individuals with amputations.

With one exception which is outlined in the subsequent paragraph, the effects of the various aspects of the NFB mechanism were as hypothesized and presented in Table I. In summary, adding a spring element promoted higher walking speeds and decreased maximum flexion angles and flexion durations. Adding friction had a minor effect on walking speed and flexion swing times, but it did aid in decreasing maximum flexion to better match able-bodied gait (55° to 65° for self-selected walking speeds [29]), and also amputee gait with high-end fluid-based systems as reported by Seroussi *et al.* [30] ($\approx 65°$), Johansson *et al.* [31] ($\approx 55°$), Blumentritt *et al.* [32] ($\approx 70°$) and Segal *et al.* [11] ($\approx 55°$ to 64°) (see Table II). Finally, as would be expected, the secondary bumper had no effect on walking speed, flexion angles, or durations, but did significantly decrease terminal impact. For flexion duration specifically, the values found in this study corresponded well with previous studies. Particularly, in the study by Boonstra *et al.*, a friction swing-phase controlled knee was compared to a pneumatically damped knee [23]. The mean flexion duration was found to be 0.54 s with the pneumatic knee performing 0.04 s better. In agreement with previous work, friction had a limited effect on decreasing swing times, whereas spring elements (extension assist) had a significant effect [5]–[7].

Conventional understanding of prosthetic gait suggests that the stiff spring systems should result in higher terminal impact [4], [7], [8]. The reasoning is that the increased knee extension force results in a more abrupt extension of the knee joint. However, based on our results this does not appear to be the case. The reason for this can be explained by considering the effect that the stiff spring has on knee flexion. The decrease in maximum flexion caused by the stiffer spring reduces the distance the shank has to travel and, therefore, allows the shank to swing with a reduced angular velocity. Particularly, angular velocities were on average 545 ($\pm$140)°/s for the Single Spring system, 536 ($\pm$97.0)°/s for the Single Stiff Spring system, 507 ($\pm$147)°/s for the Dual-Spring system and 507 ($\pm$122)°/s for the Dual Stiff Spring system. This new evidence is important as it suggests that a spring system can be tuned to concomitantly increase walking speed, provide better swing kinematics and decrease the effects of terminal impact.

In support of our secondary hypothesis, the dual-spring system outperformed the single spring system. For the same walking speed the dual-spring system resulted in lower and more normal maximum prosthetic knee flexion, decreased flexion duration, and lower terminal impact over the two walking speeds. The reduced terminal impact was achieved by the deactivation of the stiff spring and activation of the less stiff spring during the last 30° of swing before full knee extension. At that point, with adequately high friction, the shank begins to decelerate and hit the extension bumper(s) at a lower velocity. This coincides with previous work that found that even small amounts of friction at the knee axis can help to reduce the effects of terminal swing impact [33]. Providing sufficient shock absorption to amputees through the addition of specialized prosthetic components, such as shock absorbing pylons, has been shown to increase comfort, gait performance and prevent joint and back problems in the long term [34]. Therefore, the same benefits may be realized by reducing swing-phase terminal impact in prosthetic knee joints.

Fluid-based swing-phase control systems are inherently cadence responsive, automatically adjusting damping torques in response to changes in cadence to provide higher damping torques at faster speeds (see Fig. 1). It is primarily for this reason that they are indicated for active individuals, who typically adopt a variety of walking speeds. Whereas NFB systems are not cadence responsive, the dual-spring system may offer some improvement over more traditional single spring systems, by rapidly increasing torque in the later stages of knee flexion where it has a greater effect during faster walking speeds and less of an effect during slower walking speeds (see Fig. 1). The findings of this study support this hypothesis, as the dual springs outperformed the single spring by concurrently decreasing heel rise, knee flexion, and terminal impact. The dual-spring system is technically simple and, therefore, the added functional benefits can be passably justified.

The following study limitations are noted. Although the effects of swing-phase control on gait found in this study are clearly evident, a more individualized optimization of the swing-phase control variables, in contrast to the predefined settings that were used across all participants, would likely yield even better clinical results. In terms of outcome measures, future studies should include other important measurements such as gait symmetry and efficiency, as the latter would not have been feasible here given the large number of conditions that were assessed. Finally, terminal impact of existing commercial prosthesis should be quantified and subjective assessments should be used to examine the participants' perceptions of acceptable levels of terminal impact.

V. Conclusion

When properly applied and configured, NBF swing-phase control mechanisms utilizing friction and spring systems can substantially improve prosthetic gait performance for active individuals. In this study, the NFB system facilitated walking speeds that were comparable to those achieved with high-end fluid-based system. In agreement with the existing literature, the NFB systems increased walking speed, and decreased excessive prosthetic knee flexion and flexion duration. However, contrary to conventional knowledge, stiffer extension assist springs did not increase terminal swing impact. The proposed dual-spring mechanism is simple, improves prosthetic function, and offers a compact and cost effective solution for individuals around the world.

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Alex Furse received the BASc degree in mechanical engineering and the MHSc degree in clinical engineering from the University of Toronto, Toronto, ON, Canada.

He spent over two years working at Bloorview Research Institute developing prosthetic technologies. He currently works in research at Princess Margaret Hospital, part of the University Health Network, Department of Biophysics and Bioimaging, where he has been involved in development of a minimally invasive surgical tool for cancer and a portable optical device for improving wound care.

William Cleghom, photograph and biography not available at the time of publication.

Jan Andrysek received the B.Sc. (Eng.) honours degree in biological engineering from the University of Guelph, Guelph, ON, Canada, the M.A.Sc. degree in biomedical and mechanical engineering from the University of Toronto, Toronto, and the Ph.D. degree from the University of Utrecht, Netherlands.

He is a Scientist at the Bloorview Research Institute of Holland Bloorview Kids Rehabilitation Hospital in Toronto, and an Assistant Professor in the Institute of Biomaterials and Biomedical Engineering at the University of Toronto. In broad terms, his research interests relate to the study of human biomechanics and human gait, the development and evaluation of mobility assistive devices including prosthetics and orthotics, development and evaluation of quantitative gait and mobility measurement techniques, and the application of design management systems in rehabilitation engineering.