

The biomechanical effect of different spring extension assist mechanisms in mechanical 4-bar polycentric prosthetic knees for unilateral above/through-knee amputees

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Abstract

Background: Passive prosthetic knees incorporate extension assist mechanisms to enhance the swing phase of the gait cycle for persons with above/through-knee amputations. In conventional polycentric knees using a 4-bar linkage, the extension assist mechanism connects one of the bars to the body of the prosthesis. However, this design introduces a singularity in the spring action delivery, causing the spring to push the knee into flexion beyond a certain angle. In contrast, polycentric prosthetic knees with an extension assist mechanism that links the upper and lower parts of the knee eliminate the singularity feature, ensuring that the knee is pushed toward extension throughout the full range of motion.

Objective: This study aims to investigate the effects of different spring extension assist mechanisms in polycentric knees on the walking capacity of unilateral above/through-knee amputees.

Study Design: Repeated measures.

Methods: Gait data were collected at self-selected and fast speeds from 8 unilateral above/through-knee traumatic amputees using a 10-camera motion capture system. Participants trialed 2 different polycentric prosthetic knees with and without the singularity feature, both coupled with a solid-ankle-cushioned-heel prosthetic foot. They underwent gait analysis with both prosthetic knees, and a comparison analysis was conducted to examine the trial conditions.

Results: The symmetry between the prosthetic and intact knee flexion angles was higher for the design without singularity by 5.2% ($p = 0.025$) at self-selected speeds and by 7.7% ($p = 0.003$) at fast speeds. Also, peak prosthetic knee flexion angles were lower for the design without singularity by 9.4° ($p < 0.001$) at self-selected speeds and by 9.6° ($p = 0.012$) at fast speeds.

Conclusions: The extension assist mechanism without singularity improved the symmetry between the intact and prosthetic knee flexion angles by preventing excessive prosthetic knee flexion in the swing phase of the gait cycle.

Keywords

lower limb amputee, transfemoral, knee disarticulation, prosthetic design, 4-bar knee joint, extension assist mechanism, low-to-middle-income countries

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Background

The 2020 Global Burden of Disease estimated that there were 28.9 million unilateral lower limb traumatic amputees, with the highest prevalence recorded in East and South Asia, followed by Western Europe.¹ Return to movement after limb loss necessitates rigorous rehabilitation and use of prosthetic devices. Prosthetic technology has undergone significant advancements for above- and through-knee amputees, resulting in a wide range of commercially available prosthetic knees. At the advanced end of the spectrum, mono-centric microprocessor knees are available as a high-cost option. However, amputees in low- and middle-income countries (LMICs) lack financial resources to cover the cost of these state-of-art

prosthetic knees. Moreover, the design development of prosthetic devices is concentrated in European, North American, and East Asian countries and evaluated against the specifications and requirements in developed areas.² This is although most traumatic amputees live in LMICs.¹ High-cost microprocessor-controlled prosthetic knees are generally not appropriate for limited resource environments, as they often need regular battery charging, and may not be waterproof or dustproof.³

As an appropriate choice of prosthetic solutions in LMICs, mechanically controlled monocentric and polycentric prosthetic knees are commonly used. Among these options, polycentric knees offer several advantages when compared to monocentric knee joints. The instantaneous center of rotation of 4-bar polycentric

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knee joints is positioned more posteriorly and superiorly than the mechanical axes when the knee is in the fully extended position. This feature improves stability in stance and the ability to efficiently control the residual muscles, with less effort than a knee with a simple hinge mechanism.^{4,5} In addition, polycentric knees provide shank shortening and higher toe clearance in mid-swing compared to monocentric knees.^{6,7} This is beneficial as low critical toe clearance values have been associated with risks of tripping because of the small distance between the ground and toe.⁸ Therefore, there has been a growing focus on the design and use of passive polycentric mechanical knees.

During the design phase, extension assist mechanisms represent a key feature of passive polycentric knees, particularly aimed at improving the swing phase of gait.⁹ In conventional polycentric knees using a 4-bar linkage, the extension assist mechanism links one of the bars to the body of the prosthesis (Figure 1(a)). This design introduces a singularity in the delivery of the spring action: there is a specific threshold angle where the line of action of the spring aligns with the pivot point, resulting in a loss of spring action. The singularity point occurs when the link on which the spring is attached stops rotating in one direction and begins rotating in the opposite direction, whereas the knee continues its overall movement toward flexion. Conversely, extension assist mechanisms that link the bottom to the top part of the knee (Figure 1(b)) do not present this singularity feature, and the spring pushes the knee toward extension throughout the full range of motion. The singularity angle varies among polycentric prosthetic knees with this feature and is dependent on the geometry of the 4-bar linkage. The impact of the singularity on gait depends on the specific singularity angle, friction level,¹⁰ spring stiffness—with stiffer springs creating higher torque at

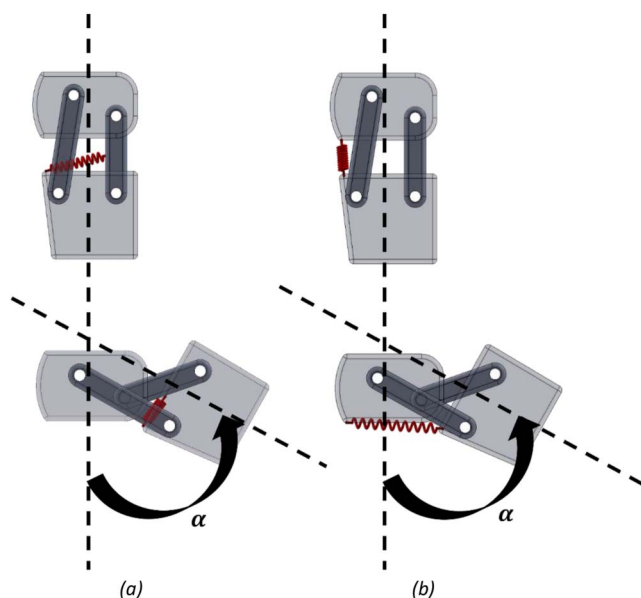


Figure 1. Prosthetic knee extension assist mechanisms. (a) Polycentric prosthetic knee that links one of the bars to the body of the prosthesis. Top: Fully extended position. Bottom: Flexed position at the singularity point where the spring pushes the knee toward extension before reaching the singularity angle (α) and toward flexion beyond that angle. (b) Polycentric prosthetic knee that links the top and bottom knee parts. Top: Fully extended position. Bottom: Flexed position; this design does not present the singularity feature, and the knee is pushed toward extension throughout the full range of motion.

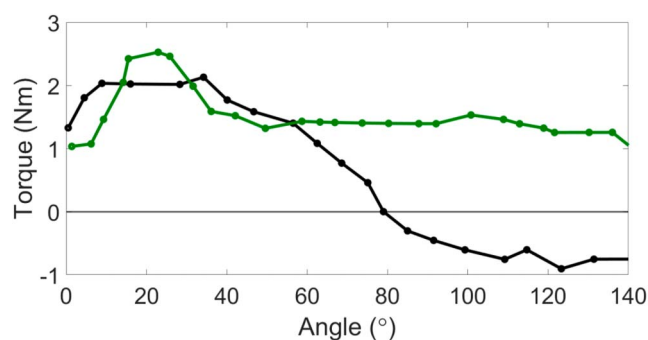


Figure 2. Torque-angle curve for a polycentric prosthetic knee with the singularity feature (black) and a polycentric prosthetic knee without the singularity feature (green).

the knee, and the spring's line of action, which changes the moment arm of the spring and, consequently, the torque profile during knee flexion. The combination of these elements is what enables optimal control in prosthetic devices.

Figure 2 presents the torque-angle curves for the polycentric prosthetic knees examined in this study, one presenting the singularity feature and one without singularity. Table 1 presents the average torque levels for motion intervals. Although these curves are specific to the prosthetic knees analyzed in this study, prosthetic knees without singularity maintain positive torque regardless of the knee flexion angle, whereas the opposite is observed for prosthetic knees with singularity. This study aims to investigate the effect of different spring extension assist mechanisms in polycentric prosthetic knees on the walking capacity of traumatic unilateral transfemoral/through-knee amputees (UTF). The hypothesis is that the extension assist mechanism without singularity improves the symmetry between the prosthetic and intact knee flexion angles, reduces the prosthetic knee flexion angles, and increases the walking speed compared to the singularity design. The outcomes of this study have the potential to offer insight for prosthetic design considerations.

Methods

Ethical approval was obtained from the Imperial College Joint Research Compliance Office and Research Ethics Committee (Reference 20IC6268). Written signed informed consent was obtained from 8 UTF healthy traumatic amputees (Table 2) who took part in the study. All participants had been prosthesis users for a minimum of 2 years and had a K3 activity level (ability or potential for ambulation with variable speed)¹¹ or higher. Seven participants had been using microprocessor-controlled prosthetic knees with

Table 1. Average torque levels throughout range of motion intervals.

Range of motion interval (°)	Average torque (Nm)	
	Singularity design	No singularity design
0–60	1.79	1.74
0–70	1.63	1.69
0–80	1.41	1.66

Table 2. Study participants details.

Participant number	Age (yrs)	Body mass without prosthesis (kg)	Height (m)	Amputation level
1	33	75.2	1.78	Above-knee
2	66	64.1	1.73	Above-knee
3	57	66.4	1.65	Above-knee
4	57	86.0	1.76	Above-knee
5	46	84.8	1.88	Above-knee
6	47	88.2	1.76	Above-knee
7	34	95.1	1.80	Through-knee
8	41	72.3	1.78	Through-knee
Mean (SD)	48 (11)	79.0 (10.4)	1.77 (0.06)	

dynamic response prosthetic feet, and one participant had been using a mechanical knee with a solid-ankle-cushioned-heel (SACH) foot.

Study participants completed walking trials in 2 different fitting cases: one polycentric knee with singularity and one without singularity, both coupled with the same SACH foot and the participant's own prosthetic socket. The order of fitting cases was randomized across participants. Although the prosthetic socket was not changed between trials, the participant's own complete shank-foot unit was removed. The unit was swapped for the new prosthesis, which used a pylon cut at the same length as the participant's prosthesis. The 2 polycentric knees were fitted with a SACH foot of the same size as the participant's own prosthetic foot. The fitting process was then followed by a familiarization period in the laboratory, where the participant spent as much time as they required to walk with the prostheses using parallel walking bars. Data collection started once the participant was comfortable walking with the prosthesis, taking half an hour on average.

Motion data were collected using a 10-camera VICON optoelectronic based motion capture analysis system (VICON

2.10.3; Oxford Metrics Group, UK). Retro-reflective markers were attached to anatomical landmarks of the lower limb: left/right anterior and posterior superior iliac spine, medial and lateral femoral epicondyles (prosthetic equivalent: medial and lateral sides of the intersection of the pyramid base plane and the pyramid center plane), medial and lateral malleoli (prosthetic equivalent: medial and lateral sides of the ankle center of rotation), second and fifth metatarsal head (prosthetic equivalent: front edge of the foot above the second and fifth toes), posterior calcaneus (prosthetic equivalent: heel at toe height), and midfoot superior (prosthetic equivalent: top of the foot, anterior to the pylon base).^{12,13} Clusters of 3 markers were placed on the thigh and shank segments.¹⁴

Each participant completed static calibration and walking trials at self-selected and fast speeds. The motion data were processed in VICON Nexus (VICON 2.10.3; Oxford Metrics Group). The gait cycle events (foot strike, toe off, and consecutive ipsilateral foot strike) were identified for each limb. The retro-reflective markers were labelled, and the marker trajectories filled using the rigid body gap filling algorithm. Marker trajectories were filtered in

Table 3. Temporal and spatial gait characteristics at self-selected (n = 8) and fast speed (n = 7).

		Singularity design, mean (SD)		No singularity design, mean (SD)	
		Prosthetic limb	Intact limb	Prosthetic limb	Intact limb
Biomechanical parameter at self-selected speed	Self-selected speed (m/s)	1.0 (0.2)		1.0 (0.2)	
	Step width (m)	0.2 (0.04)		0.2 (0.03)	
	Cadence (steps/min)	92.6 (3.9)		92.9 (8.3)	
	Stance time (% gait cycle)	56.5 (1.2) <i>p</i> < 0.001 ^a	69.4 (3.4)	58.6 (1.8)	69.1 (3.3)
	Step length (m)	0.7 (0.1)	0.6 (0.1)	0.7 (0.1)	0.6 (0.1)
	Stride length (m)	1.3 (0.2)	1.3 (0.2)	1.3 (0.1)	1.3 (0.2)
Biomechanical parameter at fast speed	Fast speed (m/s)	1.2 (0.2)		1.3 (0.2)	
	Step width (m)	0.2 (0.1)		0.2 (0.1)	
	Cadence (steps/min)	98.9 (8.1)		102.2 (7.9)	
	Stance time (% gait cycle)	54.5 (1.5) <i>p</i> = 0.002 ^a	69.6 (3.1)	56.8 (2.0)	69.1 (4.1)
	Step length (m)	0.7 (0.1) <i>p</i> = 0.015 ^a	0.7 (0.1)	0.8 (0.1)	0.7 (0.1)
	Stride length (m)	1.5 (0.2)	1.5 (0.2)	1.4 (0.2)	1.5 (0.2)

^aItalicized p values indicate statistically significant differences between prosthetic designs (*p* < 0.05).

MATLAB (The Mathworks Inc, Natick, MA) using a zero phase-lag, 4th order Butterworth filter with 6-Hz cutoff frequency.¹⁵

The absolute and nondimensional walking speeds were determined for each trial to accommodate variations in leg length among the study participants. The nondimensional walking speed (v^*) was calculated using Equation 1, where v represents the absolute walking speed, L is the leg length, and g represents gravitational acceleration.¹⁶

$$v^* = \frac{v}{\sqrt{gL}} \quad (1)$$

Inverse kinematics simulations were performed for both intact and prosthetic limbs using Freebody v2.1.^{17,18} The whole-time data series were expressed as a gait cycle percentage from 0% at heel strike to 100% at the consecutive ipsilateral heel strike, in intervals of 1% and averaged over 3 trials. Comparison analyses were performed between the 2 fitting conditions, where study participants acted as their own controls using a repeated measures design. Statistical parametric mapping was used to statistically compare the whole time-series data between prosthesis fitting conditions.¹⁹⁻²¹ IBM SPSS Statistics 28.0.0 (IBM Corp, Armonk, NY) was used to statistically compare single-valued gait characteristics. Data normality was checked using the Shapiro-Wilk statistical test. Parametric data were compared between conditions by paired t tests. Nonparametric data were compared by Wilcoxon tests. p values of less than 0.05 were considered to indicate statistical significance. The established minimal clinically important difference for walking speed is 0.2 m/s.²² However, the minimal clinically important difference for knee flexion and symmetry for UTF has not been determined.

Results

Table 3 presents the temporospatial parameters of UTF using polycentric knees with and without singularity at self-selected and fast speeds. When walking with the knee without singularity, UTF

spent a larger percentage of the prosthetic limb gait cycle in stance by 2.1% ($p < 0.001$) at self-selected speeds and by 2.3% ($p = 0.002$) at fast speeds than the singularity design. The prosthetic limb step length in fast walking conditions was higher with the no singularity design by 0.1 m ($p = 0.015$) than the singularity design. There were no significant differences in stride length, cadence, and step width between prosthesis fitting cases.

Figures 3 and 4 present the sagittal plane angles of the prosthetic and intact knee at self-selected and fast walking speeds. Statistical parametric mapping reported significantly lower prosthetic knee flexion angles at self-selected speeds for the design without singularity in stance ($p = 0.018$) and swing ($p < 0.001$, $p = 0.019$). Similarly, statistical parametric mapping showed significantly lower prosthetic knee flexion angles at fast speeds for the design without singularity in swing ($p = 0.020$). When comparing discrete values in Table 4, the peak prosthetic knee flexion angle for the design without singularity was lower by 9.4° ($p < 0.001$) at self-selected speeds and by 9.6° ($p = 0.012$) at fast speeds compared to the singularity design. The symmetry between the prosthetic and intact knee flexion angles was higher for the design without singularity by 5.2% ($p = 0.025$) at self-selected speeds and by 7.7% ($p = 0.003$) at fast speeds compared to the singularity design.

Discussion

This study investigated lower limb amputee walking speed, cadence, step length, stride length, step width, stance time, and knee flexion angles with 2 polycentric prosthetic knees with different extension assist mechanisms, one with singularity and the other without singularity to test the hypothesis that the extension assist mechanism without the singularity feature improves gait symmetry and walking speed. Conventional prosthetic polycentric knees with an extension assist mechanism that connects the main body of the prosthesis to one of the links present a singularity in the delivery of the spring action, where the knee is pushed toward

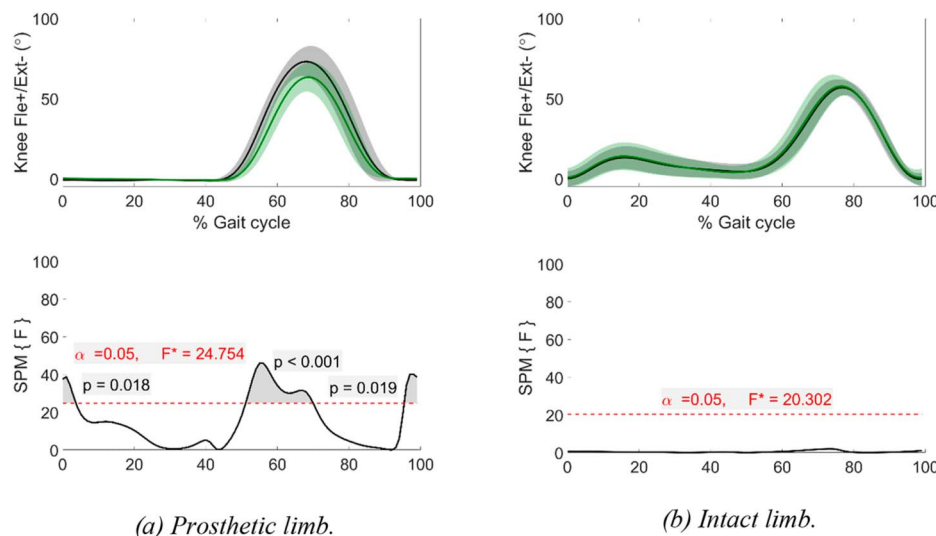


Figure 3. Top: Knee Flexion (Flex+)/Extension (Ext-) with the singularity design (black) and no singularity design (green) at self-selected speeds. Bottom: Statistical parametric mapping results ($n = 8$).

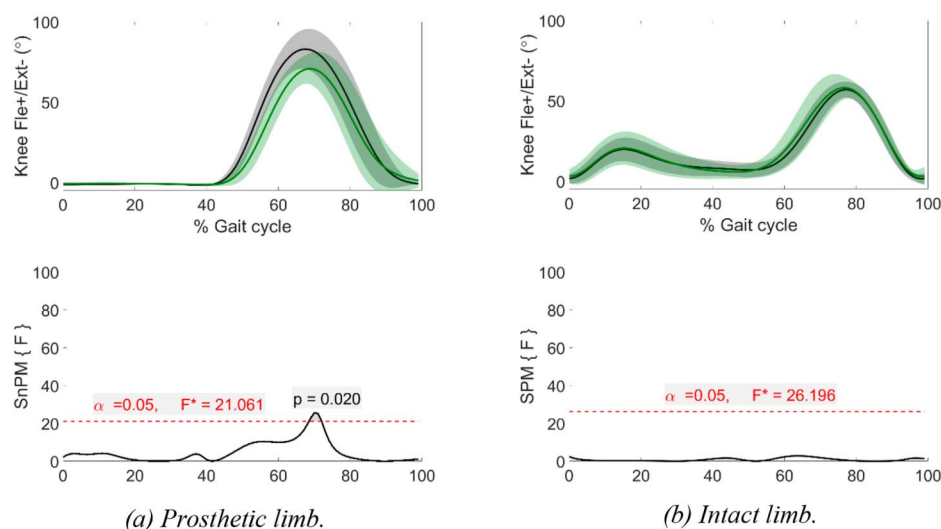


Figure 4. Top: Knee Flexion (Fle+)/Extension (Ext-) with the singularity design (black) and no singularity design (green) at fast speeds. Bottom: Statistical parametric mapping results ($n = 8$).

flexion instead of extension once it reaches a certain flexion angle (Figure 2). Conversely, prosthetic knees with an extension assist mechanism that connects the top and bottom parts of the knee do not have the singularity feature, and the spring pushes the knee toward extension across all angles. To our knowledge, this is the first study to investigate biomechanical differences between these 2 prosthetic knee categories.

The results of this study demonstrate that spring choices can improve gait symmetry for above- and through-knee amputees, confirming the hypothesis of the study. The extension assist mechanism without singularity introduced more symmetrical gait patterns than the conventional design with singularity. First, UTF spent a larger percentage of the gait cycle in the prosthetic limb stance phase with the knee without singularity than the singularity design. This indicates improved confidence and reliance on the prosthetic limb with the no singularity design, and the ability to push off the prosthetic toe for a longer time.

Second, the knee flexion angles for both speeds were significantly more symmetrical when using the knee without singularity

compared to the one with singularity. This is attributed to the extension assist mechanism without singularity, which pushed the knee toward extension and prevented excessive knee flexion in the early swing phase of the gait cycle. As a result, the peak prosthetic knee flexion was greatly reduced compared to the singularity design at both self-selected and fast walking speeds. Although this effect would seem to be beneficial (for example, by gait which looks more natural), the clinical significance of this improved symmetry is not fully known. What is known is that unsymmetrical unilateral amputee gait overloads the intact limb,²³⁻²⁵ and this overload is associated with the development of osteoarthritis.^{26,27} Therefore, it is possible that improved symmetry may improve the clinical outcome for the intact limb.

Nonetheless, the self-selected speed was similar between prostheses, indicating that the larger prosthetic knee flexion and lower symmetry observed in the design without singularity did not have a significant impact on the selected walking speeds. However, when analyzing the data collected during fast walking conditions, the results were expected to show that the extension assist

Table 4. Sagittal plane knee kinematics at self-selected ($n = 8$) and fast speed ($n = 7$).

		Singularity design, mean (SD)		No singularity design, mean (SD)	
		Prosthetic limb	Intact limb	Prosthetic limb	Intact limb
Biomechanical parameter at self-selected speed	Peak knee flexion (°)	74.2 (9.4) $p < 0.001^a$	58.5 (4.3)	64.8 (8.8)	59.5 (5.4)
	Prosthetic/intact knee flexion SI (%)	89.0 (7.7) $p = 0.025^a$		94.2 (5.1)	
Biomechanical parameter at fast speed	Peak knee flexion (°)	83.9 (12.2) $p = 0.012^a$	58.6 (4.3)	74.3 (11.4)	61.3 (5.4)
	Prosthetic/intact knee flexion SI (%)	82.8 (8.3) $p = 0.003^a$		90.5 (7.1)	

SI, symmetry index, where 100% represents perfect symmetry.

^aItalicized p values indicate statistically significant differences between prosthetic designs ($p < 0.05$).

mechanism would influence the ability to achieve higher walking speeds. As the extension assist mechanism without singularity prevented excessive prosthetic knee flexion, UTF would not have to wait for the knee to come back to the extended position from a position of higher flexion, and so would be able to walk faster, with more confidence in the safe return of the knee. Although statistical and clinical significance was not reached because of the limited sample size, a noticeable trend showed that participants walked faster with the prosthetic knee without singularity compared to the one with singularity. The analysis at the 2 different speeds may indicate that the extension assist mechanism does not significantly affect gait at normal self-selected speeds. However, it becomes evident that the singularity extension assist mechanism can potentially restrict the ability to comfortably achieve higher walking speeds and engage in more vigorous activities. It is important to note that the kinematics of the knee is also influenced by factors such as friction, spring stiffness, and spring line of action. In this study, 2 established prosthetic knees were compared without specifying the brands to prevent any conflicts of interest. The selected knees are representative of those commonly prescribed in resource-limited settings. The evaluation focused on 2 prosthetic knees featuring different extension assist mechanisms but sharing similar characteristics: both are 4-bar polycentric, mechanical, low-cost models, set up without friction and with medium spring stiffness. To ensure that the observed biomechanical differences were not influenced by friction, the design without singularity did not include any friction component, and the singularity design had its friction level set to zero. In addition, out of the 3 springs available for each knee design (low, medium, and high stiffness), the spring stiffness was set to the medium level for both knees, and no preload was used. Future studies could explore the impact of these other factors on the singularity effect.

Gait analysis alone does not fully capture the range of functions provided by the extension assist. For example, when sitting, it is crucial that the prosthetic knee remains bent. In the singularity design, the torque is close to zero between 80 and 90 degree, allowing the knee to remain bent. In the no singularity design, the extension torque is insufficient to force the knee to extend while seated, as the weight of the prosthetic foot, adaptor, and pylon exerts a flexion moment at the knee that counteracts the extension torque at the knee.

Aside from the limitation that this study looked at one design variable only and did not look at combinations of design changes across multiple variables, this study has some additional limitations. Typically, familiarization with a prosthesis takes some time, yet study participants here completed a familiarization period with the prostheses in the laboratory setting on the day of data collection. In addition, the use of microprocessor-controlled knees as the participants' day-to-day prostheses may have influenced the observed biomechanical data. However, these limitations are mitigated by providing a biomechanical comparison between 2 prosthetic knees that were both new to the participants. As the same prosthetic socket and foot were used for each participant between trials with the 2 prostheses, the observed differences between fitting conditions were because of the prosthetic knee specifications alone. The small number of participants may limit the use of these findings to a wider amputee population, and a greater sample size is required to reach strong statistical power.

Conclusion

This biomechanical study analyzed the functionality of two 4-bar polycentric prosthetic knees with different extension assist mechanisms with and without singularity. The extension assist mechanism without singularity demonstrated improved symmetry between the prosthetic and intact knee angles by limiting excessive prosthetic knee flexion during swing. Although an actively variable extension assist would be preferable to accommodate variations in walking speed, it is not currently achievable with passive devices. Therefore, there should be a specific emphasis on optimizing the design of extension assist mechanisms for polycentric knees to ensure optimal functional capabilities after amputation.

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