Gait Performance of Friction-Based Prosthetic Knee Joint Swing-Phase Controllers in Under-Resourced Settings

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Abstract: Gait quality can influence walking ability and mobility outcomes making it an important part of prosthetic rehabilitation. Prosthetic knee joint designs can influence gait quality, and limited data exists to guide component selection in under-resourced settings. This study compared spatiotemporal and kinematic gait parameters for two common types of friction-based swing-phase controlled prosthetic knee joints. Two-dimensional optical gait analysis was conducted as part of a cross-over study design involving 17 individuals with unilateral transfemoral amputations. Two prosthetic knee joints were compared. One utilized constant-friction (CF) and the other a variable cadence controller (VCC) for swing-phase control. Gait was analyzed at normal and fast walking speeds. Primary gait parameters included swing-phase time, step length, and knee flexion. Swing-phase time and peak knee flexion angles, as well as their related symmetry indices, were lower for the VCC compared to the CF ($p < 0.01$), by 11.1 to 94.1%. The VCC resulted in faster walking speeds by approximately 15% compared to the CF ($p = 0.002$). Friction-based swing-phase knee control mechanisms can facilitate an appropriate and cost-effective prosthetic knee joint solution in under-resourced settings. The findings suggest that friction-based mechanism can be designed to improve gait quality, and in turn overall walking performance.

Keywords: amputee; knee; developing country; gait analysis; prosthetic; swing-phase; swing-phase control

1. Introduction

The primary goal of lower-limb prosthetic rehabilitation is to re-establish gait and restore mobility. This is typically achieved via therapy and prescription of a prosthesis. In the case of transfemoral (TF) amputation, the prosthesis includes a prosthetic knee joint which facilitates important tasks such as sitting and walking. During typical walking gait, the knee joint provides articulation in the swing phase, while keeping the knee joint rigid through the stance phase to ensure an extended and stable limb during weight-bearing activity. A variety of knee joint mechanisms are employed clinically to achieve these functions, with varying levels of performance. Technologically advanced prosthetic knees have been shown to improve mobility outcomes via increased walking speeds and more normal movement patterns compared to simpler mechanical components [1,2]. This is important, since ambulation of individuals with TF is commonly characterized by abnormal movements which can be associated with slower, laborious gait and potential long-term

musculoskeletal problems [3–7]. Prosthetic rehabilitation outcomes are further influenced by factors such as the cause and level of amputation, age, and comorbidity [8].

In low- and middle-income countries (LMIC), simple mechanical prosthetic knee joints provide an affordable, durable and context-appropriate solution; however, this comes at the price of reduced performance compared to more advanced technologies commonly used in high-income countries [9–11]. Specifically, LMIC-relevant knee joint technologies typically comprise simple constant friction (CF) mechanisms (in some cases with a spring to promote knee extension) to control and attenuate articulation during swing-phase, thus producing more natural knee flexion and extension movements. However, CF knee joints can only achieve this at a singular walking speed, and faster gait typically results in gait deviations such as excessive swing time and heel rise (i.e., knee flexion) [12,13]. For this reason, CF knee joints are typically only prescribed for limited ambulators and elderly populations in high-income countries. Higher levels of mobility are possible via advanced prosthetic knee joint designs, such as hydraulic knees; however, these are too costly, complex and inappropriate for general use in most LMICs.

LMIC appropriate solutions have been sought that aim to improve performance while preserving LMIC appropriateness; these typically comprise simple mechanical components such as dampeners and springs. For example, the optimization of frictional damping and extension assist springs has been demonstrated in highly controlled experiments to improve swing-phase kinematics [13]. Aspects of this work ultimately led to the development of the variable cadence controller (VCC), which consists of an optimized spring in combination with a variable friction mechanism. Hence, in contrast to CF knees, the VCC has the potential to enable more natural gait patterns over a broader range of walking speeds [14]. However, to date, no formal studies have compared the gait quality of the VCC to conventional CF mechanisms. Such data could aid in elucidating the extent to which the VCC influences gait performance, and hence the potential improvements (or detriments) to gait and mobility.

The primary aim of this study was to introduce the VCC design and assess relevant spatiotemporal and kinematic gait parameters associated with swing-phase control and to compare the VCC to CF knees relevant to LMICs. It was hypothesized that the VCC would result in more symmetrical swing times, step lengths as well as reduced heel-rise (swing-phase knee flexion excursions) across different walking speeds. Secondarily, the anticipated improvements in swing-phase control were expected to improve overall walking capacity, including walking speed.

2. Methods

This study was conducted at a prosthetic facility located in Cambodia. The chosen testing location enabled access to the prosthetic components of interest (constant friction knees) used by the relevant populations (younger, highly ambulatory users). The facility was also part of a funded monitoring and evaluation project involving the deployment of knee joints using the VCC, thus allowing for a comparison of technologies. Finally, the facility had a well-established prosthetic program (providing clinical and technical services) along with some previous experience in clinical research.

2.1. Prosthetic Knee Joints

The All-Terrain knee or ATK (Legworks, Inc., Toronto, ON, Canada) has an automatically operated stance-phase lock and the VCC for swing-phase (Figures 1 and 2). The ATK is available in high-income markets and close to 20 LMICs. Studies have been conducted with the ATK to assess function, biomechanics and user satisfaction [15–18].

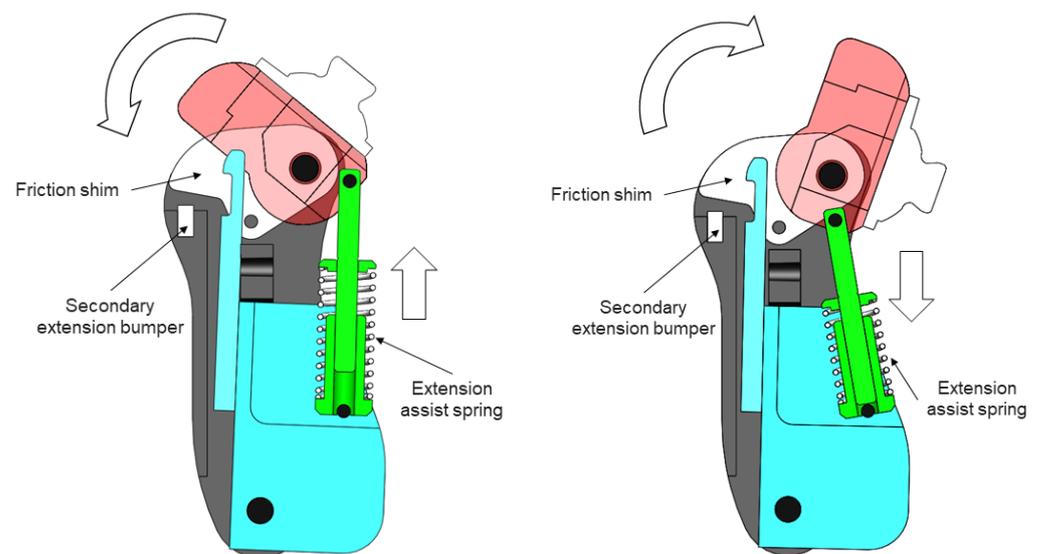**Figure 1. Left**: ICRC knee joint; **Right**: All-Terrain knee joint.

Figure 2. The main functional components of the variable cadence controller (VCC). **Left**: Increased friction and low spring force (and secondary extension bumper) decreases terminal impact near full extension; **Right**: Progressively increased friction (light pink area) and spring force with increase knee flexion helps to limit heel-rise and duration of swing-phase. The technical research details related to the development of the swing-phase controller can be found in [13,19].

The prosthetic facility where the research study was conducted used almost exclusively the International Committee of the Red Cross (ICRC) knee joint for its clientele. The ICRC knee joint (CREquipments, Coppet, Switzerland) is a single-axis manually locking knee joint with an adjustable CF swing-phase mechanism. This robust and affordable knee joint is provided as part of rehabilitation care in dozens of countries, most classified as LMIC [9].

2.2. Experimental Procedures

The study involved three separate sessions. During the first session, demographic data, participant's height, and weight were acquired. The condition of the participant's existing ICRC knee and prosthesis was assessed by a certified prosthetist and, if necessary, adjusted to ensure good working condition. One participant had loose swing-phase friction, which was corrected by tightening the designated bolt, and another participant had a loose

socket for which a sock was added. Afterwards, the process to fabricate a new prosthesis incorporating the ATK joint was initiated. Both prostheses had a polypropylene socket and Solid Ankle Cushioned Heel (SACH) foot.

Data collection session 1 was held approximately one month after the introductory session to allow time for the fabrication of the ATK prosthesis. Gait data were collected first for the ICRC knee joint. The participant was then given the ATK prosthesis to wear. Instructions and gait training (under one hour) were provided as needed to teach the participant how to operate the ATK joint during walking. Participants took the ATK knee prosthesis to use at home for one month as an acclimation period [20]. Participants returned to the facility one month later for gait data collection with the ATK.

2.3. Recruitment

A total of 17 participants were recruited and completed both sessions of the gait analysis. Eligible study participants comprised of individuals with unilateral TF amputations that were at least 18 years old, >6 months post-amputation, community ambulators, and actively using an existing prosthesis. Participants were excluded in cases of severe concomitant injury (i.e., severe traumatic brain injury, severe spinal cord injury with dysfunction or paralysis, severe burns, etc.) or having residual limb skin complications resulting in walking and balance issues. Participants were recruited from the prosthetic facility at the Department of Prosthetics & Orthotics in Phnom Penh, Cambodia. Ethics approval was obtained from the National Ethics Committee for Health Research, Cambodia (002NECHR) and Bloorview Research Institute Ethics Board (REB16-686). Participants provided written consent prior to data collection.

2.4. Data Collection

Gait data were collected using a camera-based approach based on established best practices [21,22]. Eight white round stickers with a diameter of 20 mm were adhered on the participants' right and left lower limbs over the greater trochanter, iliac crest, lateral malleolus (or in line with the intact lateral malleolus), and lateral knee axis (or over the mechanical joint center of the prosthetic knee). Sagittal plane data were collected via a digital camcorder (Canon EOS REBEL T5 W/18-55, Canon Inc., Tokyo, Japan). The camera had a minimum resolution and frame rate of 1280×720 and 25 fps, respectively. The camera was placed on a tripod at a height of 1.20 m (approximately hip height) above the ground at a distance of 3 m from the 8-m long walkway [21]. A 50 cm line was marked at the center and parallel to the walkway to allow for calibration of distances during video analysis (Figure 3). The participants were instructed to walk along the walkway at self-selected (normal) and fast-walking speeds. They walked in both directions to collect data bilaterally with two videos in each direction at each speed.

2.5. Gait Parameters

An open-source motion analysis software (Kinovea for Windows, Version 0.8.15, www.kinovea.org) and a data processing tool (MATLAB The MathWorks, Inc., Natick, MA, USA) were used to calculate gait parameters from the video data. In Kinovea, the iliac crest marker was used to calculate walking velocity (Figure 3). For the remainder of the spatiotemporal analysis, the heel of each foot was tracked in Kinovea during a gait cycle from that foot's toe-off to heel strike to calculate cadence, stride length, step length, and swing-time. Prosthetic side knee maximum knee flexion angle was based on the vectors from the knee and greater trochanter, and knee and lateral malleolus markers, respectively.

For the primary characterization of the swing-phase control performance, six parameters were calculated. Swing time indicates the timing of the swinging leg. Step length is the distance that the limb advances with each step. Maximum knee flexion quantifies the amount of heel rise. These gait parameters capture common gait deviations associated with the swing-phase of gait and prosthetic swing-phase control [12,13]. To compare differences

between the prosthetic and intact limbs, the symmetry index (SI) [23] was calculated for swing time, step length and knee maximum flexion, as follows:

$$SI = \frac{V_{prosthetic} - V_{intact}}{\frac{1}{2}\left(V_{prosthetic} + V_{intact}\right)} \times 100 \tag{1}$$

where V is the parameter measured on the prosthetic and intact limbs. Gait is regarded as symmetrical when SI ≈ 0.
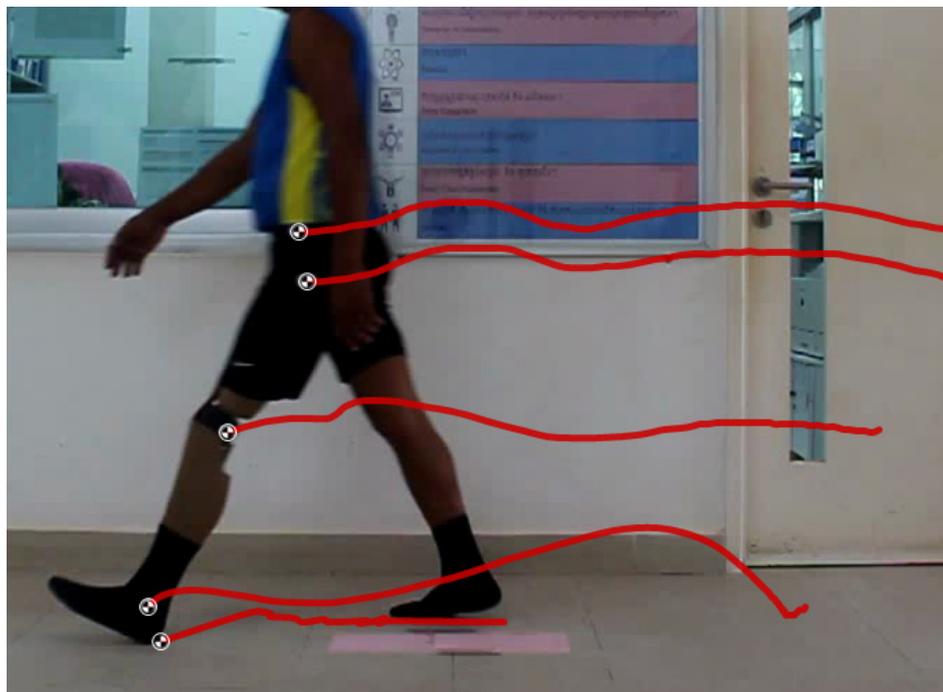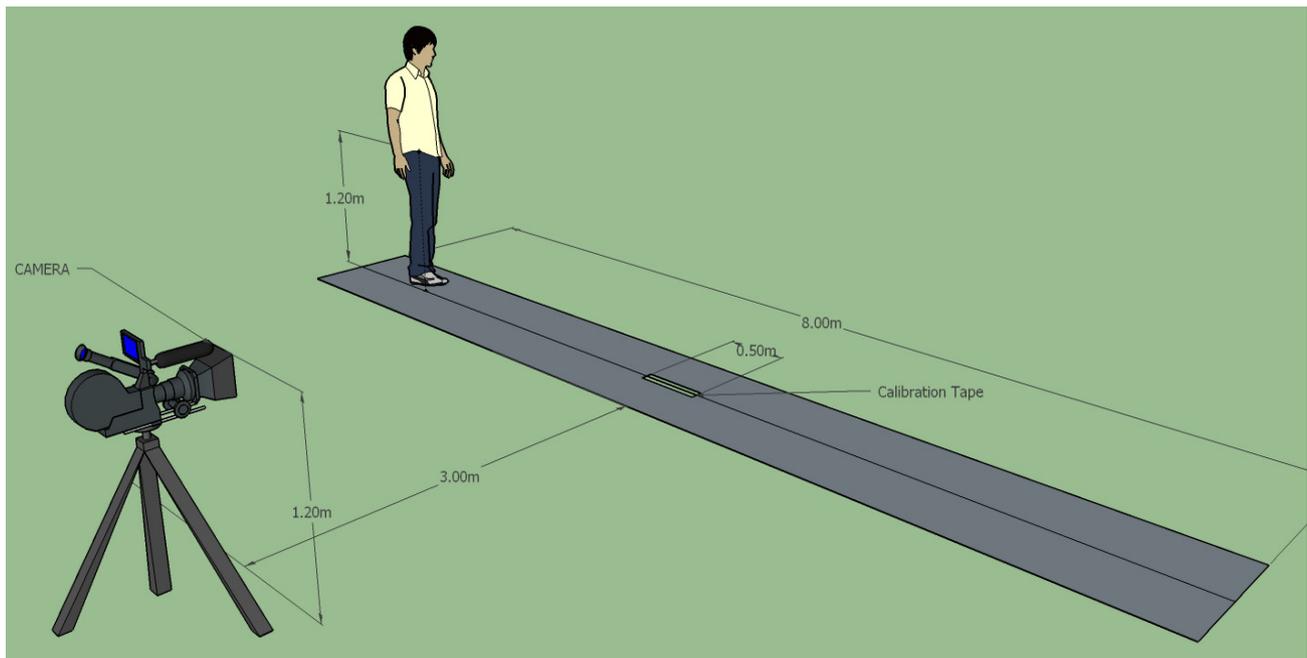

Figure 3. 2D video analysis set up (**top**) and Kinovea marker tracking (**bottom**).

Secondary measures providing a gross characterization of gait included velocity, cadence, stride length and double support time. The additional analyzed gait parameters

were chosen as they are most commonly reported in the literature and could be reliably obtained using the available gait analysis protocols [24].

Data were determined to be normally distributed using the Shapiro–Wilk test. A two-tailed paired *t*-test was applied. Data were categorized by the walking speed (NORMAL and FAST). A Bonferroni adjusted value of $p < 0.01$ was based on an initial $p < 0.05$ and primary gait parameters. Statistical analysis was performed in JMP Pro 14 (SAS Institute, Cary, NC, USA).

3. Results

Data were analyzed from the 17 participants that completed both data collection sessions. The sample comprised entirely of males, and except for one participant who lost his limb due to disease, limb loss was due to trauma (Table 1). The participants' experience in using a prosthesis ranged considerably, from very recent amputations to over 3 decades of prosthetic use. All participants were using the ICRC knee joint prior to this study.

Table 1. Participant characteristics.

Participant Descriptors	Mean $\pm$ SD, [Range]
Age (years)	40.4 ± 14.0, (20–61)
Height (m)	1.68 ± 0.07, (157–181)
Weight (kg)	63.5 ± 12.4, (42–96)
Time since amputation (years)	14.0 ± 9.8, (0.4–33)

Analysis of the primary parameters revealed significantly higher swing times for the prosthetic limb of the CF compared to the VCC. Swing times for the CF were also significantly more asymmetrical. Significant differences were also found for the prosthetic-side maximum knee flexion during the swing-phase, with the CF exhibiting greater flexion angles and asymmetry between the prosthetic and intact limbs. Step length and step length symmetry were not significantly different between the CF and VCC. These findings held across both walking speeds (Tables 2 and 3). Data for the prosthetic and intact limbs plotted in Figure 4 show the actual differences for the 3 parameters.

Table 2. Gait parameters for ICRC CF and ATK VCC for normal walking speeds.

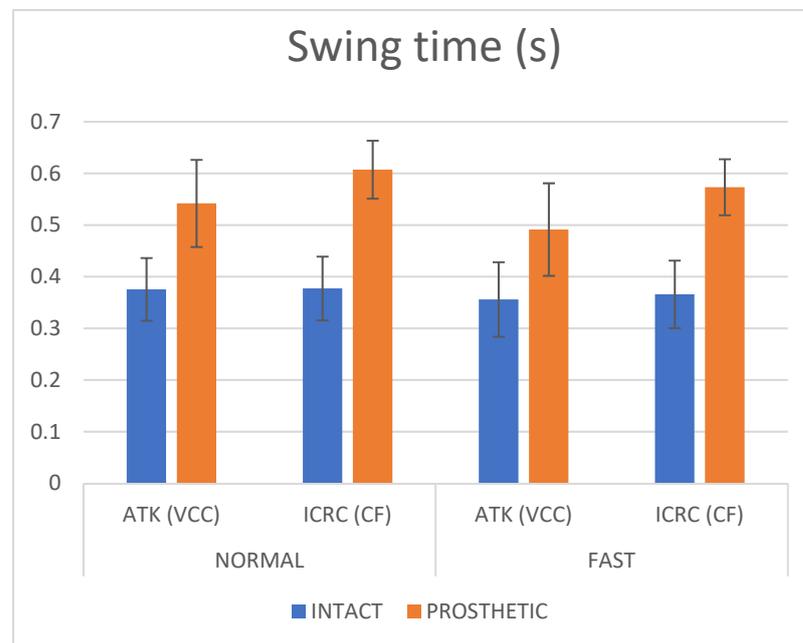| Primary Parameters	ICRC (CF)	ATK (VCC)	% Change	*p*-Value
Prosth Swing Time (s)	0.607 ± 0.010, (0.587, 0.627)	0.543 ± 0.084, (0.512, 0.571)	11.1%	**0.000**
Prosth Step length (m)	0.581 ± 0.116, (0.540, 0.621)	0.617 ± 0.111, (0.579, 0.656)	6.0%	0.094
Prosth Knee flex (deg)	72.4 ± 8.3, (68.2, 76.7)	59.9 ± 8.3, (55.7, 64.2)	18.9%	**0.000**
Swing time SI	0.48 ± 0.14, (0.43, 0.52)	0.36 ± 0.11, (0.32, 0.40)	28.6%	**0.000**
Step length SI	0.18 ± 0.22, (0.10, 0.26)	0.26 ± 0.26, (0.17, 0.36)	36.4%	0.261
Knee flex SI	0.25 ± 0.14, (0.18, 0.32)	0.09 ± 0.16, (0.01, 0.17)	94.1%	**0.002**
Secondary Parameters				
Velocity (m/s)	0.86 ± 0.09, (0.83, 0.89)	0.99 ± 0.22, (0.91, 1.07)	14.1%	**0.002**
Stride Length (m)	1.13 ± 0.10, (1.09, 1.16)	1.18 ± 0.13, (1.14, 1.23)	4.3%	0.066
Cadence (steps/min)	92 ± 7, (90, 95)	101 ± 21, (94, 109)	9.3%	0.022
Double Support (s)	0.324 ± 0.073, (0.298, 0.349)	0.302 ± 0.091, (0.270, 0.334)	7.0%	0.281

Significant differences (bolded) between ICRC (CF) and ATK (VCC) based on post hoc analysis. Mean $\pm$ one standard deviation, and 95% lower and upper confidence intervals in parentheses. SI = Symmetry Index. Percentage difference calculated as difference between the two values, divided by their mean.

Table 3. Gait parameters for ICRC and ATK for fast walking speeds.

Primary Parameters	ICRC (CF)	ATK (VCC)	% Change	*p*-Value
Prosth Swing Time (s)	0.573 ± 0.054, (0.555, 0.592)	0.491 ± 0.090, (0.460, 0.522)	15.4%	**0.000**
Prosth. Step length (m)	0.616 ± 0.135, (0.570, 0.663)	0.636 ± 0.154, (0.583, 0.690)	3.2%	0.283
Prosth Knee flex (deg)	84.7 ± 9.1, (80.1, 89.2)	68.6 ± 9.0, (64.0, 73.2)	21.0%	**0.000**
Swing time SI	0.45 ± 0.14, (0.40, 0.50)	0.32 ± 0.15, (0.27, 0.37)	33.8%	**0.000**
Step length SI	0.09 ± 0.23, (0.01, 0.17)	0.20 ± 0.27, (0.11, 0.30)	75.9%	0.022
Knee flex SI	0.39 ± 0.11, (0.26, 0.33)	0.21 ± 0.186, (0.12, 0.30)	60.0%	**0.001**
Secondary Parameters				
Velocity (m/s)	1.07 ± 0.12, (1.03, 1.11)	1.25 ± 0.32, (1.14, 1.36)	15.5%	**0.002**
Stride Length (m)	1.31 ± 0.14, (1.26, 1.35)	1.36 ± 0.14, (1.31, 1.41)	3.7%	0.103
Cadence (Steps/min)	99.6 ± 6.4, (97.4, 101)	112 ± 23.2, (104, 120)	11.7%	**0.003**
Double Support (s)	0.271 ± 0.077 (0.244, 0.297)	0.253 ± 0.088 (0.222, 0.284)	6.9%	0.370

A significant difference between ICRC (CF) and ATK (VCC) is based on post hoc analysis. Significant differences are bolded. Mean ± one standard deviation, and 95% lower and upper confidence intervals in parentheses. SI = Symmetry Index. Percentage difference calculated as difference between the two values, divided by their mean.

For the secondary gait parameters, walking velocity was significantly higher for the VCC compared to FC for both normal and fast walking conditions, while cadence was higher for the VCC during fast walking only. (Tables 2 and 3).
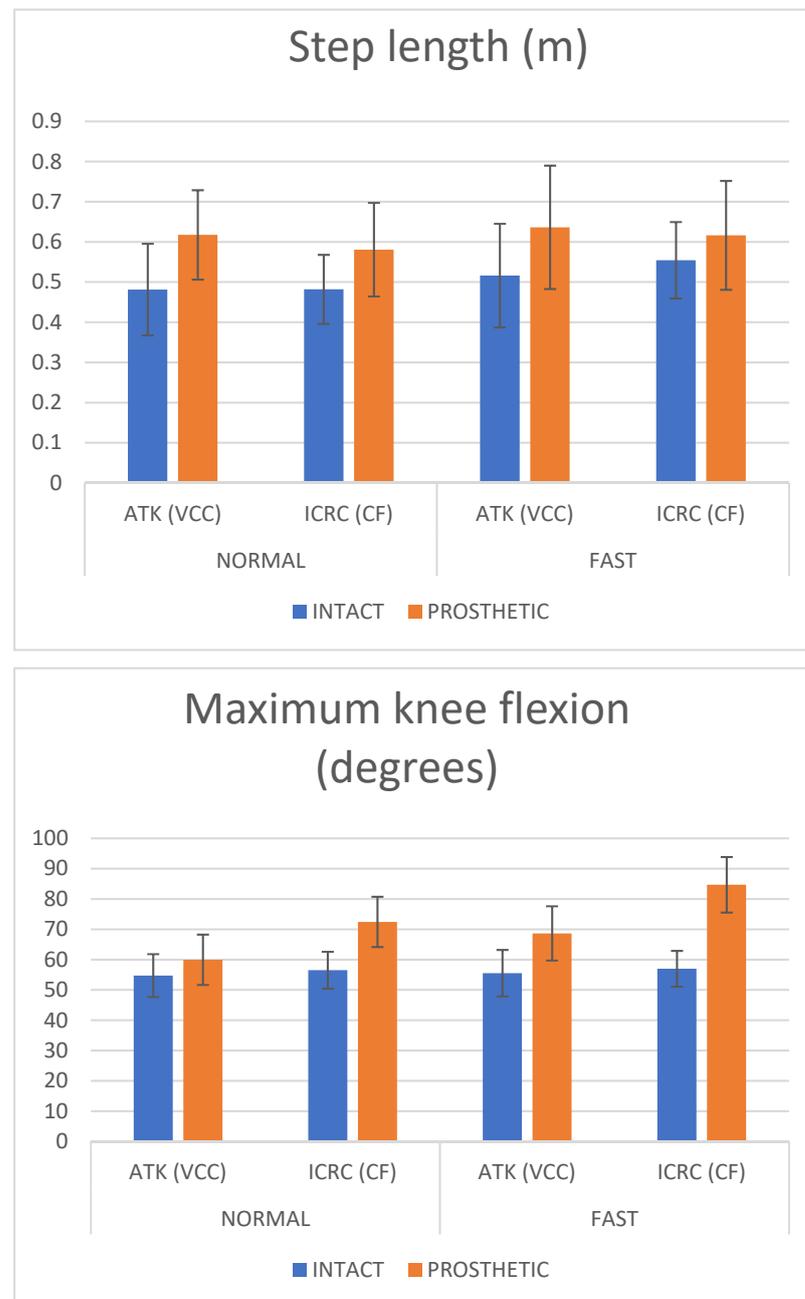

Figure 4. *Cont.*

Figure 4. Swing time, step length and knee flexion maximums for the prosthetic and intact sides for the normal and fast walking speeds. Mean values and one standard deviation shown.

4. Discussion

The goal of this work was to evaluate the gait performance associated with the VCC compared to more traditional constant friction swing-phase control knees. Both designs are based on simple and cost-effective friction braking mechanisms and, thus, generally appropriate for deployment and use in LMICs. However, a gait performance analysis of the VCC has previously not been conducted, and particularly in direct comparison with traditional constant friction mechanisms. Such data can inform clinical practices, and the potential selection of prosthetic components targeting improved clinical outcomes of prosthetic users in LMICs.

Based on the findings of this study, the VCC improves several gait patterns associated with prosthetic swing-phase control including swing-phase timing and peak knee flexion angles. Both gait parameters are reduced with the VCC, thus resulting in greater

asymmetry when compared to the intact limb. This is achieved since in contrast to constant friction mechanisms, the VCC provides a gradual increase in friction and spring force, to increasingly resist knee flexion and force knee extension to commence sooner. As such, heel rise and swing time are better controlled than with the CF.

These effects are highly consistent across the two (normal and fast) walking speeds. However, compared to the FC, the VCC produced significantly faster walking speeds at both the normal and fast walking conditions. Faster walking speeds were previously reported for the ATK compared to the constant friction knees [15,18]. The faster walking speeds are likely a result of the swing-phase control improvements associated with the VCC. This is supported by the fact that the differences in velocity between the VCC and CF were primarily due to changes in cadence rather than stride length (i.e., taking more steps rather than longer ones). However, some of the spatiotemporal and kinematic differences may also be related to differences in the stance-phase control mechanism [17]. Moreover, constant friction knees resulting in excessively prolonged prosthetic swing times and heel rise, can limit faster walking speeds, since these gait deviations become more pronounced and can disrupt normal gait biomechanics [12]. Despite walking at faster gait speeds with the VCC, the measured gait deviations (prolonged swing time and heel rise) were actually reduced compared to the CF. This supports the notion that achieving more symmetrical gait patterns can be advantageous in facilitating overall improvement in mobility and walking capacity. This is important since most individuals with transfemoral amputations exhibit slower and more fatiguing gait [3,4]. In resource-limited health care systems, which typically serve younger traumatic amputee populations capable of utilizing and benefiting from higher performance componentry, the VCC may be considered a viable option. However, it is important to further study and consider other factors such as technological appropriateness, cost, durability, ease of use, and cultural preferences amongst others.

While the ATK improved several aspects of gait related to swing-phase resulting in more symmetrical gait patterns, substantial differences (asymmetries) between the intact and prosthetic side gait parameters remained, as seen in Figure 3. These differences may in part be due to limitations associated with friction-based systems, in contrast to hydraulic or microprocessor knee joints that provide more precise swing-phase control over a broader range of conditions. Secondly, while instructions were provided for setting up the VCC, it is unknown how well the prosthetists were able to follow these instructions, and therefore whether the optimal setup and adjustments of the VCC were achieved. Nevertheless, this represents the realities associated with the deployment and use of prosthetic technology in LMICs, and therefore the data are more likely to represent the actual performance of both knees in these settings. It further speaks to the importance of technology that is intuitive and simple to use by prosthetists and prosthesis wearers alike.

This study has several limitations. Since it was conducted in an LMIC, the collection was limited by the available gait analysis technology. Video-based gait analysis was performed using standard optical recording equipment. While such systems can provide valid and reliable data [25], they do have limitations in terms of accurate capture of certain parameters [22]. Hence, only sagittal plane gait parameters were acquired. Due to the resolution, lighting, and low frame rates, markers were blurry in some frames, and this limited the accurate extraction of certain gait parameters, such as ankle and hip joint angles, where markers are near each other, and small positional inaccuracies result in large errors in segmental angles. Hence, these parameters were excluded. All participants were long-time users of the ICRC knee, and it is unclear whether the acclimation time was adequate to facilitate motor relearning, and the optimal utilization of the VCC knee joint. Nevertheless, the highly significant gait differences found between the knee joints suggest that acclimation was likely achieved.

5. Conclusions

This study provides new evidence about the differences in walking patterns related to knee joints used in LMICs. The ATK VCC swing-phase mechanism resulted in gait

improvements compared to the ICRC constant friction knee joint. In particular, the variable resistance during swing-phase provided by the VCC resulted in reduced and more symmetrical swing-phase timing and knee flexion angles. Faster walking speeds were also measured for the VCC, which may be associated with the VCC's improved swing-phase control performance. The findings should be considered in the development and provision of prosthetic knee joint components in LMICs where the need for technology that is affordable, durable, and functional persists.

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