Differences in Knee Flexion between the Genium and C-Leg Microprocessor Knees While Walking on Level Ground and Ramps

Derek J. Lura, PhD1, Matthew M. Wernke, PhD2, Stephanie L. Carey, PhD2, Jason T. Kahle, MSMS, CPO3, Rebecca M. Miro, PhD3,4, M. Jason Highsmith, PhD, DPT, CP3,4

1. Florida Gulf Coast University, Department of Bioengineering and Software Engineering, Fort Myers, FL
2. University of South Florida, Mechanical Engineering Department, Tampa, FL
3. University of South Florida, School of Physical Therapy and Rehabilitation Sciences, Tampa, FL
4. University of South Florida, Center for Neuromusculoskeletal Research, Tampa, FL

Corresponding Author and Reprint Requests:
Derek J. Lura, PhD
Florida Gulf Coast University
Department of Bioengineering and Software Engineering
U.A. Whitaker College of Engineering
10501 FGCU Boulevard South
Fort Myers, FL 33965-6565
dlura@fgcu.edu
(239) 590-7832 office
(239) 590-7304 fax

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Abstract

Background: Microprocessor knees have improved the gait and functional abilities of persons with transfemoral amputation. The Genium prosthetic knee offers an advanced sensor and control system designed to decrease impairment by: allowing greater stance phase flexion, easing transitions between gait phases, and compensating for changes in terrain. The aim of this study was to determine differences between the knee flexion angle of persons using the Genium knee, the C-Leg knee, and non-amputee controls; and to evaluate the impact the prostheses on gait and level of impairment of the user.

Methods: This study used a randomized experimental crossover of persons with transfemoral amputation using the Genium and C-Leg microprocessor knees, with an observational sample of non-amputee controls. Gait analysis by 3D motion tracking of subjects ambulating at different speeds on level ground and on 5° and 10° ramps was completed.

Findings: Use of the Genium resulted in a significant increase in peak knee flexion for swing (5°, p<0.01, d=0.34) and stance (2°, p<0.01, d=0.19) phases relative to C-Leg use. There was a high degree of variability between subjects and significant differences still remain between the Genium group and the control group’s knee flexion angles for most speeds and slopes.

Interpretation: The Genium knee generally increases flexion in swing and stance, potentially decreasing the level of impairment for persons with transfemoral amputation. This study demonstrates functional differences between the C-Leg and Genium knees to help prosthetists determine if the Genium will provide functional benefits to individual patients.

<Abstract Word Count: 245/250>
1. Introduction

Biomechanical analysis has been widely utilized to evaluate the performance of lower limb prosthetic components. In a review of measures and procedures to determine value of microprocessor knees, joint angles were used in 5 of the 37 included papers [1]. Another 2013 review found 3 of 27 relevant articles assessed knee flexion during stance when peak stance and swing knee flexion angles for the C-Leg (Otto Bock Healthcare, Duderstadt, Germany) and Mauch SNS (Ossur, Reykjavik, Iceland) knees when walking at a controlled speed [3]. There was not a significant difference between the knees during stance flexion, but there was a significant increase in peak swing flexion for the Mauch SNS possibly due to less swing damping [3]. Related gait changes included significantly ($p<0.05$) increased self-selected walking speed and sound side step length. During stance phase, the prosthetic knees both had less peak flexion than the intact limb, and both had greater flexion than the intact limb during swing phase. Kaufman et al. also evaluated stance phase knee flexion of the C-Leg relative to non-microprocessor controlled knees in a non-randomized cross-over study and found a significant increase ($p<0.05$) in stance flexion between C-Leg and non-microprocessor knees [4].

Recently, Bellmann et al. [5] evaluated knee kinematics in a sample of 9 persons while using 4 different microprocessor knee systems: C-Leg, Rheo Knee (Ossur. Reykjavik, Iceland), Adaptive 2 Knee (Chas A. Blatchford and Sons. Hampshire, United Kingdom) and the Hybrid Knee (Natebsco. Tokyo, Japan). Authors stated two important aspects of microprocessor mediated swing phase control; maximal knee flexion angle (which should correspond to typical gait at 60-65°) and comfortable (as opposed to abrupt) terminal knee extension. Briefly, increased walking speed tended to increase peak knee flexion in swing phase. Across the four knee systems, the range of swing phase peak flexion increases per corresponding speeds were $11.7\pm4.9^\circ/m/s$ when using C-Leg compared to $22.3\pm9.1^\circ/m/s$ while using the Hybrid knee. The rate at which knee extension was terminated was also variable between knee systems. For example, the Hybrid knee stopped knee extension
abruptly, the Adaptive 2 decelerated knee extension discontinuously, and the Rheo provided
the most gradual knee extension cessation.

The previous investigation [5] using four knee systems reported C-Leg as being the most
consistent in terms of swing phase knee flexion angle relative to gait speed. However, an
investigation comparing C-Leg with Genium revealed that Genium far improves peak swing
knee flexion consistency when walking on flat ground [6]. Authors also reported that when
walking with small steps, knee flexion angle is significantly increased, and more typical
compared to the C-Leg. While the C-Leg has reportedly improved knee kinematics for many
persons with transfemoral amputation (TFA), these early comparative effectiveness studies
suggest the Genium system may further normalize knee movement. However, these
observations have not been studied on varied terrains and also need to be verified on flat
ground.

The purpose of this study was to determine if peak stance and swing knee flexion angles
would be greater when persons with transfemoral amputation use the Genium
microprocessor knee system compared with the C-Leg regardless of terrain or walking
speed. A second purpose was to determine kinematic differences between persons with TFA
and non-amputee controls while using the Genium knee relative to the C-Leg.

2. Methods
All procedures were reviewed and approved by the University of South Florida's Institutional
Review Board. The study was listed on clinicaltrials.gov (#NCT01473992), and subjects
gave informed consent prior to participation in the study.

2.1. Study Design Overview
This study was a randomized experimental crossover of TFAs using the Genium and C-Leg
microprocessor knees and included observational measures of a non-amputee control
sample. Twenty TFA and five control subjects ($n=25$) participated in the study. Each TFA
subject tested on both knee conditions, in a random order separated by an accommodation
period of $\geq 2$ weeks to $\leq 3$ months, depending upon when subjects determined their readiness.
to test. Subjects’ gait was recorded using passive reflective markers and an 8 camera Vicon (Oxford, UK) motion analysis system. Subjects walked on flat ground with and without ankle weights at very slow, slow, normal, and fast walking speed conditions, and without weights on a 5’ and 10’ ramps at slow, normal, and fast walking speed conditions.

2.2. Randomization and Interventions

All subjects entering the study were C-Leg users for at least one year prior to enrollment in the study. An electronic random number generator was use to assign subjects to either continue with their C-Leg or to be fitted with a Genium at their time of recruitment and consent to participate. The study prosthetist was notified of the subject’s assigned condition via telephone on the day of the subject’s knee fitting. All fittings and adjustments were performed by the study prosthetist who was state-licensed and certified by the American Board for Certification in Orthotics, Prosthetics and Pedorthics, as well as by Otto Bock Healthcare for fitting both C-Leg and Genium MPK systems. The subjects’ prosthetic sockets and suspension systems were not changed for the duration of the experiment to reduce the confounding effects from fit and acclimation issues. All subjects were fit with an Otto Bock Trias (standard height) or Axtion (low profile) foot, based on limb length, for use over the duration of the study. Manufacturer specifications were used to set componentry alignment, and were verified using the LASAR (Otto Bock) alignment system.

2.3. Fitting and Accommodation Periods

After enrollment, anthropometric data, randomization order, and study foot were recorded. Knee fittings and alignment were conducted and recorded. All subjects were invited to return to the study prosthetist or physical therapist for adjustment, alignment and training at their discretion as many times as they wished to optimize their fit, comfort and function and to mirror real clinical practice and component prescription. These visits were counted and reasons for each visit were recorded. Subjects received training from the study physical therapist on the functions of both knee systems for transitional movements, obstacle
crossing, ramps, stairs, speed variation, and variable surfaces. Portions of the training
techniques used in this protocol have been previously published [7, 8]. The minimum accommodation period was 2 weeks, after which subjects were contacted
weekly to determine their ability to walk without personal assistance on 1) level ground, 2) inclines, 3) declines, 4) up & down stairs, and 5) on uneven ground. Subjects could contact investigators at any time after the 2 week minimum to declare their readiness to physically
demonstrate that they had accommodated to their currently assigned knee and study foot. Subjects were considered accommodated after verbally acknowledging and physically
demonstrating their ability to ambulate independently on all 5 of the previous conditions. This study accommodation test was adapted from Hafner et al. [9]. Following accommodation, subjects were scheduled for A-phase testing. Following A-phase testing, knee units were
switched and the process was repeated culminating in B-phase testing.

2.4. Testing

Passive marker based stereo photographic motion tracking was selected to record and
analyze knee movement because it is valid, reliable and considered to be the gold-standard
measurement technique for gait analysis [10]. Specifically, an 8 camera Vicon motion
analysis system was used to collect knee kinematic data of subjects performing gait tasks.
Passive reflective makers were attached to subjects using a combination of neoprene straps
and double side adhesive collars. A description of each marker is given in Table 1. AMTI
(Watertown, MA) force platforms were embedded in the gait platform and ramp, and used to
record heel strike and toe off, the subsequent heel strike was recoded with kinematic
approximation [11]. In accordance with manufacturer specifications and recommendations,
the Vicon cameras were calibrated and force platforms zeroed before each session and after
adjusting the ramp to the designated slope condition.
Table 1: Description of motion analysis marker set.

<table>
<thead>
<tr>
<th>Name</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>RBAK</td>
<td>Middle spine of the right scapula</td>
</tr>
<tr>
<td>CLAV</td>
<td>Jugular notch</td>
</tr>
<tr>
<td>STRN</td>
<td>Xiphoid Process</td>
</tr>
<tr>
<td>T1</td>
<td>1\textsuperscript{st} Thoracic Vertebra</td>
</tr>
<tr>
<td>T10</td>
<td>10\textsuperscript{th} Thoracic Vertebra</td>
</tr>
<tr>
<td>R/LASI</td>
<td>Anterior superior iliac spine of the pelvis</td>
</tr>
<tr>
<td>R/LPSI</td>
<td>Posterior superior iliac spine of the pelvis</td>
</tr>
<tr>
<td>R/LIC</td>
<td>Medial crest of the ilium</td>
</tr>
<tr>
<td>R/LGT</td>
<td>Greater trochanter</td>
</tr>
<tr>
<td>R/LTH1-4</td>
<td>Thigh cluster markers</td>
</tr>
<tr>
<td>R/LLK</td>
<td>Lateral epicondyle of the femur</td>
</tr>
<tr>
<td>R/LMK</td>
<td>Medial epicondyle of the femur</td>
</tr>
<tr>
<td>R/LSK1-4</td>
<td>Shank cluster markers</td>
</tr>
<tr>
<td>R/LLA</td>
<td>Lateral malleolus of the ankle</td>
</tr>
<tr>
<td>R/LMA</td>
<td>Medial malleolus of the ankle</td>
</tr>
<tr>
<td>R/LTOE</td>
<td>Distal head of the 2\textsuperscript{nd} metatarsal</td>
</tr>
<tr>
<td>R/LHEE</td>
<td>Calcaneus at the same height as LTOE while standing</td>
</tr>
</tbody>
</table>

All subjects completed the tasks in the same order. Four walking speeds started with normal self-selected speed on level ground, then proceeded to slow (i.e. casual), then very-slow (i.e. deep in thought, or admiring an object), and fast (i.e. hurried, late for a plane) speeds. The weighted walking condition followed with a 0.5kg ankle weight attached to each leg to simulate heavy footwear. The 5\textdegree ramp test was completed at normal, slow, and then fast speeds prior to the 10\textdegree ramp test in the same order. Subjects were given the opportunity to rest between each test. A total of 20 different trial types were assessed and each type was completed twice to record data over the force platform for both the prosthetic and anatomical leg.

2.5. Data Processing

Tracking segments were defined using the surface markers and a large amount of redundancy was included in the marker set to compensate for marker dropout, and to increase data consistency and reliability. Frames were tracked by least squares minimization of the in-segment marker reconstruction error [12]. Transformations between segments were then used to estimate the anatomical joint centers using a spherical best fit method [13]. A combination of anatomical markers and calculated joint centers were then used to define
anatomical segments. Segment axes definitions were based on the recommendations of the International Society of Biomechanics [14]. Joint angles were calculated from Euler angle transformations of anatomical segments.

Although the marker set used enabled tracking of the entire lower body, knee flexion was the measure of interest that addressed our a priori hypotheses, and therefore was the only measure included in this report. The gait cycle was defined from heel strike to heel strike of the involved foot. Peak knee flexion during stance phase was defined as the maximum knee flexion angle from 0 to 30% of the gait cycle to assure peak stance flexion in the loading response. Peak swing flexion was defined as the maximum knee flexion angle from 30 to 100% of the gait cycle. These definitions were used to prevent subjects who used the ‘ride-down’ strategy on the ramp decent from inflating the stance flexion angles after the loading response, and further increasing deviations in subject data.

2.6. Statistical Analyses

Statistical analyses were performed with IBM SPSS (Armonk, NY). Data were compiled into a database, assessed for completeness and descriptive analyses were performed (i.e. means, standard deviations). The Shapiro-Wilk test was used to determine if data were normally distributed. Normally distributed data were assessed using univariate analysis of variance (ANOVA) for both peak stance and swing flexion angles by knee type (Control, Genium Dominant, C-Leg Dominant, Genium Prosthetic, and C-Leg Prosthetic) and trial type as fixed factors to determine global differences between knees, followed by a Tukey’s post-hoc multiple comparison analysis. When data were abnormally distributed, non-parametric equivalent tests were used. The a priori level of statistical significance was ps0.05. Finally, investigators assumed that any missing data would be missing totally at random and adopted the “last observation carried forward” or “next observation carried backward” method as the study’s a priori intention to treat plan [15, 16].
3. Results

3.1. Subject Demographics

Complete data were collected from all 25 consented subjects. The 20 unilateral TFA subjects repeated the data collection (once on C-Leg and once on Genium). Left and right legs for the control subjects were both considered to be part of the same condition, and the left and right leg of the amputee subjects were classified into dominant (non-amputated) or prosthetic leg. The majority of TFA subjects were male (80%) with a mean (SD) age of 46.5(±14.2) years and BMI of 26.4(±4.2) kg/m². The majority were employed (55%), 25% were governmentally classified as “disabled” and the remaining 20% were students or retired. All TFA subjects were independent, unlimited community ambulators (Medicare Functional Level 3). The mean time since amputation was 17.7(±15.6) years and amputation etiology was predominantly trauma (70%) followed by malignancy (20%), and peripheral vascular disease (10%). Mean relative residual limb length was 70(±30)% of the sound side femur and the mean hip flexion contracture angle was 12.8(±7.7)° as measured with a manual goniometer in the Thomas Test position. The majority of subjects (70%) had the dominant leg amputated. A variety of prosthetic sockets (i.e. ischial ramus containment, ischial support, subischial, quadrilateral) and suspension systems (i.e. locking liners, suction, elevated vacuum) were utilized by amputee subjects.

The five non-amputee controls included three males and two females. Control subjects’ mean age was 57.2(±15.7) years and BMI was 23.0(±3.0) kg/m². Three control subjects were employed and two were retired. Control subjects were free from health maladies.

3.2. Overall

Results for between categories effects for the stance and swing peak flexion resulted in significant differences (p<0.01) between knee types, trial types, and knee & trial combined. Partial Eta Squared was calculated in SPSS to estimate the effect size for differences between knee conditions including controls, and was large ($\eta_p^2=0.29$) for swing phase and ($\eta_p^2=0.59$) stance phase. The effect size of differences between the prosthetic knee
conditions only was evaluated by Cohen’s $d$ [17], and was approaching a medium magnitude
effect ($d=0.34$) for swing phase, and was small ($d=0.19$) for stance phase. Tukey post-hoc
analysis of swing phase peak knee flexion revealed significant differences ($p<0.01$) between
both Genium prosthetic side ($55°±14$) and C-Leg prosthetic side ($51°±14$) relative to each
other, and to controls ($65°±8$). Differences for swing flexion between dominant legs were not
significant between to either knee condition, or controls. Prosthetic side stance phase peak
flexion angles were significantly different ($p<0.01$) between control ($25°±11$), Genium
prosthetic side ($7°±10$), and C-Leg prosthetic side ($5°±9$). Dominant side stance flexion was
significantly different ($p<0.01$) between control and C-Leg condition ($22°±5$), and between
control and Genium condition ($23°±7$, $p=0.03$), but not between between Genium and C-Leg
($p=0.13$). Descriptive statistics for peak knee flexion of each task for the prosthetic knee
conditions and controls are given in Table 2. Amputee data were abnormally distributed, thus
a Wilcoxon Signed Rank test was used to compare means between knee conditions for each
task.
Table 2: Average (SD) swing and stance phase peak knee flexion in degrees.

<table>
<thead>
<tr>
<th>Trial Type</th>
<th>Swing Phase</th>
<th>Stance Phase</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Control</td>
<td>Genium</td>
</tr>
<tr>
<td>Very Slow</td>
<td>54 (5)</td>
<td>48† (8)</td>
</tr>
<tr>
<td>Slow</td>
<td>59 (4)</td>
<td>55† (5)</td>
</tr>
<tr>
<td>Normal</td>
<td>64 (3)</td>
<td>62† (6)</td>
</tr>
<tr>
<td>Fast</td>
<td>69 (5)</td>
<td>64 (5)</td>
</tr>
<tr>
<td>Very Slow</td>
<td>56 (5)</td>
<td>49* (8)</td>
</tr>
<tr>
<td>Slow</td>
<td>59 (5)</td>
<td>55* (6)</td>
</tr>
<tr>
<td>Normal</td>
<td>63 (4)</td>
<td>60* (6)</td>
</tr>
<tr>
<td>Fast</td>
<td>69 (6)</td>
<td>65* (5)</td>
</tr>
<tr>
<td>Slow</td>
<td>57 (6)</td>
<td>48* (12)</td>
</tr>
<tr>
<td>Normal</td>
<td>62 (4)</td>
<td>56† (8)</td>
</tr>
<tr>
<td>Fast</td>
<td>67 (4)</td>
<td>57* (14)</td>
</tr>
<tr>
<td>Slow</td>
<td>62 (5)</td>
<td>54 (13)</td>
</tr>
<tr>
<td>Normal</td>
<td>68 (4)</td>
<td>63† (6)</td>
</tr>
<tr>
<td>Fast</td>
<td>72 (6)</td>
<td>60 (14)</td>
</tr>
<tr>
<td>Slow</td>
<td>58 (7)</td>
<td>37 (17)</td>
</tr>
<tr>
<td>Normal</td>
<td>65 (4)</td>
<td>47† (17)</td>
</tr>
<tr>
<td>Fast</td>
<td>66 (4)</td>
<td>49 (18)</td>
</tr>
<tr>
<td>Slow</td>
<td>69 (4)</td>
<td>60 (17)</td>
</tr>
<tr>
<td>Normal</td>
<td>76 (4)</td>
<td>60 (16)</td>
</tr>
<tr>
<td>Fast</td>
<td>81 (6)</td>
<td>58 (18)</td>
</tr>
<tr>
<td>Total</td>
<td>65 (8)</td>
<td>55† (14)</td>
</tr>
</tbody>
</table>

Statistically significant differences between Genium and C-Leg indicated by *(p<0.05) or †(p<0.01)

3.3. Walk

For overground walking, Genium knee use resulted in an average swing phase peak knee flexion increase of 5-7° (p≤0.05) for all walking speeds except fast walking. Fast walking with Genium resulted in a 2° increase in peak swing flexion that was not significantly different from C-Leg. During stance phase, like swing phase, Genium use resulted in 1-2° greater peak flexion at all speeds. Only slow (p=0.01) and normal (p=0.02) walking speeds reached the a priori level of statistical significance. No significant differences were observed in stance or swing phase for the dominant leg between the two knee conditions. Generally, knee flexion angles resulting from Genium use tended to be closer to control peak knee angles than those resulting from knee than the C-Leg use, but angles for the Genium knee were still
less than controls for all prosthetic side conditions. See Table 2 and Figure 1 for average prosthetic side flexion angles and standard deviations.

![Graphs showing knee flexion angles for different conditions](image)

**Figure 1:** Prosthetic side peak knee flexion on flat ground. Statistically significant differences between Genium and C-Leg indicated by *(p<0.05)* or †(p<0.01).

### 3.4. Weighted Walk

Swing phase peak knee flexion with Genium use (prosthetic side) was found to be significantly greater *(p<0.02)* by 3-6° for all speeds during the weighted walk trials. During stance phase, Genium use at very slow *(p=0.05)* and fast *(p<0.01)* speeds were significantly increased by 2-3° for the prosthetic side, and the normal *(p=0.03)* speed was significantly increased by 3° for the dominant side. Both prosthetic knees performed similarly for weighted and unweighted walking whereby larger flexion angles were observed for Genium compared with C-Leg, but were still less than control for all prosthetic side conditions. Knee flexion seemed to be more impacted by the gait speed than the additional weight.

### 3.5. 5° Ramp

Genium use resulted in significantly more swing phase knee flexion on the prosthetic side for 5° ramp ascent for slow *(8°, p=0.02)*, medium *(7°, p<0.01)*, and fast *(3°, p=0.01)* speeds. For descent, only the normal speed was significantly increased *(9°, p<0.01)* during swing phase.

No significant differences were observed between prosthetic knees on the involved side for
stance phase ascent. For stance phase decent slow (3°, \(p=0.04\)) and fast (4°, \(p=0.04\)) speeds Genium use significantly increased stance flexion peaks relative to C-Leg on the involved side. Both prosthetic knees produced very small amounts of stance flexion during ramp ascent compared to controls and the dominant legs. Though not reported here, this resulted in greater stride asymmetry than observed during level walking. The Genium knee was found to produce more flexion than the C-Leg in all cases except the swing phase of the fast decent. Swing phase peak flexion angles for 5° and 10° ramps are given in Figure 2, and for Stance phase in Figure 3.

![Graphs showing knee flexion angles during ramp ascent and descent](image)

**Figure 2:** Swing phase prosthetic side peak knee flexion on ramps. Statistically significant differences between Genium and C-Leg indicated by *(p<0.05)* or †(p<0.01).

### 3.6. 10° Ramp

Similar results were observed on the 10° ramp relative to the 5° ramp, with even more pronounced asymmetry during ramp ascent. There was a very large deviation in angles, especially on the prosthetic side. The only significant difference observed was increased involved side knee flexion of the Genium Knee relative to C-Leg for the swing phase of ramp ascent at subjects’ normal pace (8°, \(p<0.01\)). Further, 10° ramp ascent appeared to be the most stressful task tested as this resulted in the smallest stance flexion angles recorded in the study.
4. Discussion

Due to engineering provisions for increased flexion angles with the Genium knee system, we
hypothesized that Genium would generally increase peak flexion events in stance and swing
regardless of terrain condition, compared to the C-Leg. This was the case in the great
majority of conditions tested. Previous studies identified C-Leg as improving peak swing
knee flexion symmetry compared with non-microprocessor systems [3]. Peak (early) stance
flexion improved by 2.3°, however the knees were still in hyperextension and the
improvement was not statistically significant [3]. Segal et al. concluded there were minimal
biomechanical differences between C-Leg and the non-microprocessor comparator and
perhaps more challenging activities may highlight differences. Where Segal et al. identified
minimal kinematic differences; Kaufman et al. reported a more natural loading response with
the C-Leg compared to non-microprocessor knee systems. C-Leg use was reported to result
in a statistically significant (p<0.01) 4-5° increase into a flexed posture during loading
response and an internal knee extension moment [4]. These studies demonstrate that the C-
Leg microprocessor knee system made improvements in gait biomechanics compared to
non-microprocessor systems, but there was still room for improvement.
More recently, clinicians have begun to speculate whether consistency of knee angle at multiple speeds may be a factor limiting the walking performance of the person with transfemoral amputation. Bellman et al. demonstrated that compared with other MPK systems, the C-Leg best stabilized the swing flexion angle compared with the Rheo, Adaptive 2, and the Hybrid MPK systems [6]. Bellman et al. went on to compare peak swing knee flexion between the C-Leg and Genium knees at multiple walking speeds. In the Bellman study, the C-Leg varied swing knee flexion over approximately 10° (≈57-66°) whereas the Genium maintained flexion consistently within two degrees, between 62-64° [6].

In our study, we found comparably high consistency of peak swing flexion angle at normal and fast speeds (≤3°) regardless of terrain condition. The greatest difference in normal and fast speed peak swing knee flexion angles were in weighted walking to simulate heavy footwear such as boots where knee angle varied 5°. The slow walking and very slow walking speeds introduced greater challenges in maintaining peak swing flexion consistency. Across all speeds and conditions, swing flexion consistency ranged from 2-16° with Genium use. Including the very slow speed in the testing protocol enabled the ability to observe this trend.

With slow and very slow walking, potentially when mentally focusing on things other than walking, a higher degree of stability is probably useful. Thus, having the feet on the ground more than in the air is partly facilitated by having a smaller knee flexion angle [18]. Bellman et al. [6] reported that even during slow walking the Genium had higher knee flexion angles than the C-Leg. This agrees with our findings, however, we found that Genium produced higher peak swing flexion angles even with very slow walking as well as during up and downhill and weighted walking. Bellman et al. [6] further reported that when making small steps, likely the more comparable test to the very slow walking condition in this study, the C-Leg failed to initiate swing phase 24.7% of the time compared with Genium that failed to initiate swing only 4.9% of the time. This improved swing initiation may partially explain the difference in peak swing angles between the knee systems.

In stance phase, the Genium has a trend of being in a more flexed posture relative to the C-Leg regardless of condition tested. However, the improvements are not as pronounced.
during the swing phase of some conditions, such as the 10° ramp ascent, suggesting prosthetic designs can be further improved. On flat ground, the Genium significantly improved peak stance flexion at slow and normal speeds where previous studies showed the C-Leg as improving stance flexion over non-microprocessor knees. At extreme speeds (i.e. very slow and fast) however, peak stance flexion was not significantly different between knees. When walking downhill, TFAs could flex the prosthesis far greater than when walking uphill, regardless of speed. The C-Leg improved downhill walking speed and quality compared to NMPKs [19], however this study shows the Genium offers increased improvements in the knee movement pattern relative to the C-Leg for downhill walking. Uphill walking is still an area requiring greater attention in the rehabilitation of persons with TFA as prosthetic knees are only negligibly flexed under this condition.

4.1. Limitations and Future Research

All subjects' knee software settings and alignments were conducted by the same prosthetist and to the satisfaction of the team of study prosthetist, study physical therapist and subject. Accommodation and training was provided and independent ambulation on all terrains assessed prior to data collection. Thus, confounding factors from software and alignment settings were minimized. The study is limited by the use of surface markers that are subject to error in placement and soft tissue deformations and movement. Motion between the residual limb and socket interface, and deformation of residual limb soft tissues were not characterized. These uncharacterized motions may lead to additional variance in knee angle calculations, since the biomechanical model used assumed that the thigh was a rigid segment. This likely contributed to the large standard deviations observed, especially during fast walking and ramp trials which were more dynamic and more likely to produce tissue deformations. Future study should consider kinetics, subject stability and actual gait speeds at the different inclines and speeds conditions.
5. Conclusions

Accommodation, training and use of the Genium microprocessor prosthesis was found to produce increased knee flexion compared with the C-Leg in both the stance and swing phases of gait. This increased knee flexion is clinically significant as it better recreates a normalized, anatomic movement pattern. The knee flexion angle of the non-amputated leg was not significantly affected by use of the Genium relative to the C-leg. Control subjects typically had the greatest knee flexion, followed by the amputees’ sound side, and then prosthetic side of the subjects with the Genium and C-Leg knees respectively. This shows that Genium use increases stance and swing knee flexion angles compared with the C-Leg, but improvements are still possible, especially in certain walking conditions such as when walking uphill.

6. Acknowledgements

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