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The Effects of Increased Prosthetic Ankle Motions on the Gait of Persons with Bilateral Transtibial Amputations

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Abstract

Objective: To determine if the provision of prosthetic ankle motion improves walking performance in persons with bilateral transtibial amputations.

Design: Cross-over experimental design in which nineteen persons with bilateral transtibial amputations were fitted with Endolite Multiflex Ankles (flexion unit) and Otto Bock Torsion Adapters (torsion unit) to increase relative motion between the prosthetic foot and socket in the sagittal and transverse planes, respectively. Quantitative gait analyses were performed on subjects as they walked with four prosthetic configurations: baseline without flexion or torsion units, with only the flexion unit, with only the torsion unit, and with both the flexion and torsion units. Data were compared with a control group of fourteen able-bodied subjects.

Results: The flexion unit increased ankle sagittal plane motion (6°-7°) and increased positive ankle power (about 0.17 watt/kg). The torsion unit increased transverse plane ankle range of motion by 1°-2°. Responses from questionnaires indicated that 14 of the 19 subjects preferred the prosthetic configuration that included both the flexion and torsion units. Further, the subjects perceived that the

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increased prosthetic ankle motion was particularly beneficial for improving stability while they walked on uneven terrain.

Conclusions: Both the subjective and objective results suggest that prosthetic foot and ankle components that allow for greater sagittal and transverse plane rotations provide substantial benefit during walking and should be considered for persons with bilateral transtibial amputations. Nonetheless, clinicians should perform individual and appropriate assessments of patients to insure that they are capable of using components that may improve mobility while possibly sacrificing some degree of stability.

Keywords

Ankle; Gait; Kinematic; Kinetic; Prosthesis; Transtibial Amputation

The inferior walking performance of persons with amputations may, at least in part, be attributed to the current state of prosthetics technology. More functional prosthetic components may produce significant improvements in users' walking performance¹. Many studies have analyzed and compared the gait of persons with unilateral transtibial amputations walking with different prosthetic feet and ankles²⁻⁷. However, persons with unilateral transtibial amputations often exhibit compensatory actions on the sound side and display asymmetrical joint motions and forces^{4, 8-11}, making it difficult to assess prosthetic performance or identify specific deficiencies in their function.

The anatomical foot and ankle form a complex structure with the important functions of providing stability, efficient forward progression and shock absorption during walking. During stance phase, three rockers of the foot and ankle¹² act in a serial fashion to allow the body to advance forward over the stance leg. Ankle plantarflexion during the loading response phase of gait facilitates foot flat to ensure stability as the leg is progressively loaded, and the ankle simultaneously provides shock absorption through eccentric contraction of the ankle dorsiflexors. Additionally, the compliance and motion provided by the anatomical ankle joint enables more of the plantar surface of the foot to come into contact with the walking surface, particularly on uneven terrains, thus enhancing stability.

Some investigators have studied how different prosthetic foot types affect unilateral amputee gait, but few have specifically investigated prosthetic ankle mechanisms. Wirta et al.¹³ compared the effects of five ankle-foot devices on the gait of transtibial amputees and found that the amputees preferred those devices that developed less shock and had greater damping at heel strike. van Jaarsveld et al.¹⁴ felt that the declination of peak accelerations at heel strike was an important aspect of foot-ankle performance in transtibial amputees. Perry et al.⁷ recommended that prosthetic designs should provide amputees with improved ankle mobility that attempts to capture the dynamic characteristics of a normal articulation between the foot and shank segments during the early stance weight acceptance period. These studies indicate that the amount of actual or simulated prosthetic ankle motion can appreciably change actual and perceived walking performance in unilateral amputee gait.

The purpose of this study was to determine if the provision of prosthetic ankle motion in persons with bilateral transtibial amputations significantly improves their walking performance. Studying people with bilateral amputations offers a unique opportunity to observe prosthetic gait because they generally ambulate with greater right-left symmetry¹⁵ than unilateral amputees and they do not have the ability to compensate for the prosthesis with an intact limb. If the provision of prosthetic ankle motion improves stability, efficient forward progression and shock absorption during walking, then we would expect to observe subjects to walk at faster freely-selected speeds and with increased prosthetic ankle power in late stance as a result

of higher energy storage and return. Additionally, with the incorporation of components that improve prosthetic function we would anticipate greater user satisfaction.

METHODS

Subjects and prosthetic configurations

The subjects were recruited from clinics and prosthetics fitting centers in the Chicago metropolitan area. Criteria for inclusion were specified as individuals who were a minimum of two years post-amputation; used prostheses as their primary means of mobility; and were without serious health issues that would directly affect gait. Inclusion in the study was not limited by age, weight, height or residual limb length. All subjects who met these inclusion criteria were recruited by their physicians or prosthetists and asked to participate in our study. Subjects signed consent forms that were approved by Northwestern University's Institutional Review Board.

The study consisted of four phases. At the beginning of each phase subjects were fitted with a different prosthetic configuration that they used for a minimum of two weeks, concluding with a data collection session. A quantitative gait analysis was performed at the end of each phase to record the walking pattern of subjects using the different configurations.

For phase I, an experienced, certified prosthetist fitted all subjects with Seattle Lightfoot II feet (**baseline** configuration) having appropriate keel stiffness that was selected based upon their weight and activity level. Subjects continued using their existing socket and suspension type for the duration of the study. No other prosthetic component was controlled by the researchers. The Seattle Lightfoot II uses a Delrin keel and is claimed by the manufacturer to provide fatigue resistance, durability, spring-like resistance, and shock absorption for a smooth, dynamic gait. For each of the different prosthetic configurations, the bench alignment of the prostheses was based upon the manufacturer's recommendations for the Seattle Lightfoot II. The prostheses were then fitted on the subjects and dynamic alignment was performed at the discretion of the prosthetist as subjects walked in the laboratory. After two weeks' accommodation to the prosthetic feet, subjects participated in a quantitative gait analysis.

Phase II began immediately upon completion of the first gait analysis session. In addition to the Seattle Lightfoot II feet, subjects were fitted bilaterally with either Endolite Multiflex Ankles (**flexion** configuration) or Otto Bock Torsion Adapters (**torsion** configuration). Selection of the ankle components for phase II was randomly determined. During phase III, each subject received the alternate components to those assigned in phase II. In the final phase, both the Endolite Multiflex Ankles and the Otto Bock Torsion Adapters (**combined** configuration) were fitted into the subject's prostheses.

Endolite Multiflex Ankles primarily provide plantarflexion/dorsiflexion movement and allow a small degree of inversion/eversion rotation. The stiffness of the Endolite Multiflex Ankles is controlled by combining different durometer rubber balls and snubbers, which are selected by the prosthetist based upon the weight and activity level of the user. The unit is claimed by the manufacturer to facilitate ambulation on slopes more naturally and with less effort, to adapt more easily on uneven ground, and to provide smooth transition from plantarflexion to dorsiflexion during walking. The Otto Bock Torsion Adapter provides up to 20° of internal and external rotation. The prosthetist adjusted the torque resistance to meet the user's needs and preference by turning a screw located in the center of the torsion adapter. Torsion adapters are believed to reduce the shear stress between the socket and the residual limb, providing increased comfort during gait¹⁶. This reduction in transverse plane shear may enable users to turn more naturally with greater fluidity.

A single prosthetist fitted all of the subjects with the prosthetic components for this study. The prosthetist confirmed that components were adjusted to best suit the needs of the subject, and insured that the subject felt safe walking with the new prosthetic configuration. The subjects were initially fit with the resistance of the torsion components at the manufacturer's neutral setting. As the individuals walked, the resistance of the torsion components were decreased until transverse plane motion was apparent to the prosthetist. He then asked the subject if that was a comfortable position of resistance. If not, he incrementally increased the resistance back to a position which felt more stable to the subject. Proper alignment of the prosthesis is important because it may allow for a more energy efficient gait and an optimum distribution of pressures within the socket. Misalignment may decrease the functions provided by the components, and may also increase the effort required to walk.

Gait Data Acquisition

Data collection and analyses for the study were conducted in the VA Chicago Motion Analysis Research Laboratory (VACMARL). The VACMARL has an eight-camera Eagle Digital Real-Time motion capture system (Motion Analysis Corporation [MAC], Santa Rosa, CA) that is used to measure marker movements. A modified Helen Hayes marker set¹⁷ was used to define a biomechanical model of the participant. As the subject walked along the walkway, the positions of the markers were recorded by the motion analysis cameras mounted around the periphery of the room. Six force platforms (Advanced Mechanical Technology, Inc [AMTI], Watertown, MA) located midway along the walkway and embedded flush with the floor were used to measure ground reaction forces (GRFs). Both the kinematic and kinetic data were collected using EVa Real Time Software (EVaRT, MAC, Santa Rosa, CA). The kinematic data were acquired at 120 Hz and the kinetic data were simultaneously recorded at a sampling rate of 960 Hz. The GRF and motion data were used to calculate joint moment and power via inverse dynamics using OrthoTrak software (MAC, Santa Rosa, CA).

During the gait analyses, the subjects were instructed to ambulate at their comfortable, freely-selected walking speed, then they walked at their fastest comfortable speed, and finally at their slowest comfortable speed. A total of 10 to 15 trials of data were collected for each walking speed and the subjects were given the opportunity to rest at any time during the experiment.

Surveys that incorporated a 5-point Likert scale were used to document the subjective perceptions of the participants toward the various components, and were administered following each gait analysis. Four surveys—one for each prosthetic configuration—were developed with similar questions, but with a few statements specific to the function of each prosthetic configuration. Each questionnaire consisted of a number of statements that asked subjects to rank from one to five their subjective assessment for each particular configuration. A value of one indicated “I strongly disagree,” three was neutral, and five indicated “I strongly agree.” The baseline questionnaire simply recorded subjective perceptions to walking with the Seattle Lightfoot II Feet and without any ankle units. The other questionnaires prompted subjects to compare the flexion, torsion and combined configurations to the baseline configuration. Additional comments that captured subjective perceptions about each configuration were also documented. After completing the study, the research subjects were given the option of keeping some or all of the prosthetic components (Seattle Lightfoot II Feet, Endolite Multiflex Ankles, Otto Bock Torsion Adapters) that had been fitted in their prostheses.

Data processing and analysis

Missing data points were interpolated with a cubic spline technique. The raw data were then filtered using a fourth-order bi-directional Butterworth infinite-impulse response digital filter having an effective cut-off frequency of 6.0 Hz. OrthoTrak (MAC) software was used to calculate temporal-spatial data, joint angles, GRFs, joint moments, and powers. The mean

values of parameters of interest were computed for each subject and then averaged across all subjects.

Temporal-spatial, kinematic, kinetic variables and questionnaire results were examined using statistical methods. The following parameters were compared between the groups: walking speed, step length, cadence, step width, stance time, double support time, ankle plantarflexion/dorsiflexion range of motion (ROM) during stance phase, ankle transverse plane rotation ROM, knee flexion/extension ROM in stance phase, knee flexion/extension ROM during a gait cycle, hip flexion/extension ROM, pelvic obliquity ROM, magnitude of the first peak of the vertical GRF, peak fore-aft acceleration GRF, peak fore-aft deceleration GRF, peak ankle plantarflexion moment, peak ankle dorsiflexion moment, peak positive ankle power, peak negative ankle power, peak positive hip power, and peak negative hip power. Statistical analysis of the gait analysis data utilized a repeated measures ANOVA to test the differences among the four prosthetic configurations. When main effects were found to be significant at a level of $p < 0.05$, pairwise comparisons were made using Bonferroni adjustments for multiple comparisons. Values for p were adjusted by the software to reflect the Bonferroni correction. Data from 14 able-bodied persons walking at self-selected slow speeds that are kept on file in a laboratory database were used to provide comparisons with the amputee subjects. The control group consisted of 7 men and 7 women. Their averaged age, height, and mass were 26 years, 174.2cm, and 72.3kg, respectively. Independent t -tests were used to investigate differences between the different prosthetic configurations and able-bodied data. The questionnaire data were analyzed using non-parametric statistics. A Friedman test was used to determine if responses to the different questions were significantly different at $p < 0.05$. Where significant differences existed, pairwise comparisons between the different interventions were analyzed using a Wilcoxon signed-rank test with level of significance adjusted using a Bonferroni correction.

RESULTS

A total of 19 bilateral transtibial amputee subjects were enrolled in the study. The average age of the subjects was 52.8 years (S.D. $\pm$ 17.6 years, Table 1). Their average height and mass were 171.9cm and 77.4kg, respectively. The amputation etiology for 10 of the subjects was due to trauma, and for the other 9 was secondary to peripheral vascular disease (PVD)¹⁸.

Data collections were performed at three different walking speeds. However, the different prosthetic interventions were determined to have no appreciable effect on walking speed. Therefore, only the data that were collected at the subjects' freely-selected speed will be reported. For analysis, these data from the research subjects were compared with data from able-bodied control subjects walking at their slowest comfortable speed (0.82 m/sec).

During the gait analyses, four subjects used a single point cane to assist walking. While ambulating, these four subjects always held the cane in the right hand and in contact with the ground during the left stance phase. The results showed that all subjects, including the four that used canes, demonstrated reasonably good symmetry during gait and similar vertical GRF magnitudes were observed for both legs. Nonetheless, to avoid potential artifact from the use of a cane, only the right side data from subjects were analyzed. Those four subjects who used canes did not seem to rely appreciably on their cane to support their body weight. They probably used the cane to provide a sense of security, to improve stability and to prevent falling rather than to provide significant load bearing during walking.

Temporal-spatial data

Among the four configurations tested, no significant changes were observed in freely-selected walking speed, step length, cadence, stance phase duration (as a percentage of the gait cycle),

and double support time (Table 2). When walking with the four prosthetic configurations, the amputee subjects displayed a significantly higher cadence compared to the able-bodied subjects. When the subjects walked with the **baseline** configuration (only the Seattle Lightfoot II), they displayed wider step widths compared to when they walked with the **flexion** configuration (with Endolite Multiflex Ankles, $p=0.009$), the **torsion** configuration (with Otto Bock Torsion Adapters, $p=0.046$), and the **combined** configuration (with both Endolite Multiflex Ankles and Otto Bock Torsion Adapters, $p=0.006$). Additionally, the amputee subjects displayed wider step widths when walking with baseline ($p=0.04$) and torsion ($p=0.024$) configurations compared to the able-bodied individuals.

Gait kinematics

Results from the analysis of kinematic parameters are reported in Table 3. The subjects had over 6 degrees more ankle sagittal plane ROM when the flexion unit was included in the prosthesis (Table 3, $p<0.001$ for baseline vs. flexion, baseline vs. combined, flexion vs. torsion, torsion vs. combined). Compared to able-bodied controls, the amputee subjects displayed smaller ankle sagittal plane ROM when walking with baseline and torsion configurations ($p<0.001$ for both comparisons). The ankle sagittal plane ROM was not statistically different between the able-bodied subjects and the amputee subjects walking with flexion and combined configurations. When the subjects walked with the flexion unit, the plantarflexion angle was increased during loading response phase and the dorsiflexion angle also was greater at late stance phase (Figure 1). When the subjects walked with the torsion unit, they demonstrated about 2 degrees greater ankle transverse plane ROM (Table 3, Figure 2, $p=0.026$ for baseline vs. torsion, $p=0.002$ for flexion vs. torsion, $p=0.014$ for baseline vs. combined, and $p<0.001$ for torsion vs. combined). The amputee subjects' ankle transverse plane ROM when walking with all four prosthetic configurations was smaller than that of able-bodied subjects ($p<0.001$). Similarly, the amputee subjects displayed less stance phase knee flexion and greater hip sagittal plane ROM for all of the prosthetic configurations compared to the able-bodied subjects.

Gait kinetics

Results from analysis of the kinetic data are presented in Table 4.

The magnitude of the first peak of the vertical GRF was significantly greater when the subjects walked with the combined configuration than with the baseline configuration ($p=0.023$, Figure 3). The peak acceleration fore-aft GRF was greater when the flexion unit was included ($p=0.023$ for baseline vs. flexion, $p=0.029$ for baseline vs. combined, $p=0.036$ for flexion vs. torsion, and $p=0.002$ for torsion vs. combined, Figure 4). The peak deceleration fore-aft GRF had a greater magnitude when the subjects walked with the flexion unit ($p=0.038$ for baseline vs. flexion, $p=0.005$ for baseline vs. combined, $p=0.019$ for flexion vs. torsion, and $p=0.001$ for torsion vs. combined, Figure 4). The peak ankle plantarflexion and dorsiflexion moments were decreased when the subjects walked with the flexion unit (Figure 5) compared to when they walked without it. However, the decrease in ankle moments was not statistically significant. When walking with the four prosthetic configurations, the amputee subjects consistently displayed smaller ankle plantarflexion moments and greater ankle dorsiflexion moments compared to the able-bodied subjects. The peak ankle power absorption had a greater magnitude when the subjects walked with flexion configuration, but it was statistically significant only when compared with the torsion configuration ($p=0.017$, Figure 6). When the flexion unit was added to the prosthesis, the subjects displayed increased peak ankle "power generation" at the end of stance phase ($p<0.001$ for baseline vs. flexion, $p<0.001$ for baseline vs. combined, $p<0.001$ for flexion vs. torsion, and $p=0.003$ for torsion vs. combined, Figure 6). The peak ankle "power generation" was smaller when the amputee subjects walked with the four prosthetic configurations compared to the able-bodied subjects, even when the flexion unit was included. The peak hip power absorption during late stance phase had a greater

magnitude when subjects walked with the combined prosthetic configuration than with the baseline configuration ($p=0.009$, Figure 7). The magnitudes of both the peak positive and negative hip power were greater for the amputee subjects walking with all four prosthetic configurations than they were for the able-bodied subjects.

Questionnaires

The questionnaire results are presented in Table 5.

Subjects indicated that they perceived they were able to walk longer distances using the combined configuration than they could with the baseline and torsion configurations ($p=0.002$ and $p=0.005$, respectively, Table 5). They also reported that walking was smoother with the flexion configuration than with the baseline configuration ($p=0.007$). Compared to the baseline configuration, subjects reported that the torsion configuration enabled them to turn more easily ($p=0.002$). Subjects perceived that it was easier to walk on uneven ground using the flexion, torsion, and combined configurations rather than with the baseline configuration ($p=0.001$, $p=0.004$, and $p=0.002$, respectively).

After the 2nd, 3rd and 4th gait analyses subjects were asked which prosthetic configuration they liked best. After the 2nd gait analysis, all subjects preferred the components that had been added for the 2nd phase of the study (either the torsion or flexion unit) compared to the baseline condition. After the 3rd gait analysis, all of the subjects had walked on the baseline, torsion, and flexion configurations; 8 subjects preferred the torsion configuration, 10 subjects preferred the flexion configuration, and one subject liked both configurations equally well. After the 4th gait analysis, 14 subjects indicated that they preferred the combined configuration, 3 subjects preferred the flexion configuration, and 2 subjects liked the torsion configuration best. Those subjects who did not prefer the combined configuration responded that they felt unstable, especially during standing, because either the flexion or the torsion unit provided too much motion. After walking with each prosthetic configuration, subjects were also asked to express other perceptions about their experience that were not necessarily captured in the Likert scale statements (Table 6).

DISCUSSION

The freely-selected walking speed is commonly measured and used as an indicator of walking performance. Although the results of this study indicated that the freely-selected walking speeds with the flexion, torsion, and combined configurations were slightly greater than those in the baseline configuration, the different prosthetic configurations did not produce significant changes to freely-selected walking speed. A majority of studies that compared dynamic response (DR) feet with the SACH (solid ankle cushioned heel) feet in persons with unilateral transtibial amputation reported a tendency toward increases in freely-selected walking speed, but those increases were not statistically significant either^{2-6, 19-21}. Therefore, freely-selected walking speed in lower-limb amputees may be affected by other aspects of the prosthesis other than prosthetic foot or ankle function.

In this study, the different prosthetic configurations with the ankle components did not significantly affect the cadence or step length of persons with bilateral transtibial amputations. Previous studies of unilateral transtibial amputees have generally reported no significant differences in cadence and step length when comparing the gaits of subjects walking with a DR foot and a SACH feet^{2-6, 19-21}. Only one investigation reported a significant increase in cadence when using a DR foot compared to the SACH feet³. In this current study we observed increased step length when subjects walked with the flexion unit compared with the baseline configuration, but the results were not statistically significant.

An increased step width in lower-limb amputees is often attributed to decreased stability during walking and a poorer perception of balance. Persons with better walking performance generally ambulate with narrower step width, which has been associated with improved walking efficiency²². Step width of the bilateral amputee subjects was observed to be about 50% wider than that of the able-bodied controls during gait. When subjects walked with the flexion, torsion and combined configurations, they had significantly narrower step widths compared to the baseline configuration. This finding indicates that providing greater prosthetic ankle sagittal and transverse plane motion during late stance may have improved their dynamic stability during ambulation, or at least made the subjects feel more stable. Further studies are required to determine exactly how prosthetic ankle components affect standing and walking balance.

When subjects walked with the flexion unit, their sagittal plane ankle range of motion significantly increased compared to the baseline and torsion configurations (Table 2 and Figure 1). The ankle joint motion with the flexion unit was comparable to that of the able-bodied individuals, differing by only about 1-2°. The altered ankle motions may be explained by the mechanical properties of the prosthetic feet and ankle components and how they differ from the anatomical counterparts of able-bodied individuals, particularly by the lack of active muscle controlling the ankle joint and the absence of active plantarflexion in late stance phase. Reduced stiffness in the flexion unit of the prosthesis enabled subjects to walk with a comparable peak-to-peak sagittal plane ankle motion (Figure 1), but it may also have decreased balance and stability during standing. Further investigations are needed to determine the proper stiffness required for prosthetic ankle components and feet and relate those values to stability and range of motion.

In able-bodied persons, the ankle plantarflexion motion that occurs during the loading response phase of walking assists in absorbing shock as load is rapidly transferred from the trailing to the leading leg¹². Increased ankle plantarflexion was observed in the bilateral transtibial amputees when the flexion unit was added, which apparently provided a better shock absorption mechanism and increased comfort during ambulation. The increased ankle plantarflexion that was provided during early stance may also have improved the user's walking stability by reducing the time required to reach foot flat and begin progressing the center of pressure forward under the foot⁷. The increased ankle dorsiflexion motion in amputees during terminal stance restored some portion of the ankle and forefoot rocker mechanisms, which are believed to contribute to an improved roll-over and thereby facilitate forward progression of the body¹². Roll-over shapes²³ of the foot and ankle were not investigated for this study, but they should be in future investigations because they would undoubtedly be affected by prosthetic ankle motion.

A statistically significant increase in transverse plane ankle rotation ROM was observed when subjects walked with the torsion unit, but the amount of increase was only about 2° on average (Table 3). The torsion unit enabled subjects to achieve a total of about 5° peak-to-peak transverse plane ankle rotation during straight, level walking, but that amount was still significantly less than the approximately 17° measured in the able-bodied controls (Figure 2). Additionally, the transtibial amputee subjects in this study exhibited very different transverse rotation motion patterns at the ankle than that demonstrated by able-bodied persons. One possible explanation for the reduced motion in the amputee subjects is that the stiffness setting of the torsion adaptor was simply set too high. Alternatively, the subjects may have felt unstable walking with greater ankle transverse rotation so they could have adopted a walking pattern to avoid increased rotation. Regardless, it is still unknown whether transverse plane rotation on the order of that utilized by able-bodied ambulators would be of benefit for prosthesis users. The questionnaire results indicated that subjects felt the torsion unit was particularly beneficial for turning and walking on uneven ground. Gait analysis of subjects as they walk on different

floor conditions or perform tasks like turning²⁴ and walking up/down stairs may illustrate further the effects of the increased prosthetic ankle motions.

The bilateral transtibial amputee subjects displayed a significantly greater magnitude in the first peak of the vertical GRFs when they walked with the combined configuration compared to baseline configuration (Table 4, Figure 3). However, there were no significant differences between the baseline configuration and the torsion or flexion configurations. The magnitude of the first peak in the vertical GRF is generally known to increase with walking speed. The difference in the first peak vertical GRFs between the combined and the baseline configurations in this study could be due in part to slightly increased walking speed with the combined configuration. However, in this current study, it is not known if the increased GRF peak can be attributed to changes in the ipsilateral prosthesis, the contralateral prosthesis (affecting the transition from one leg to the other), or a combination of both. Studies that investigated the walking performance of persons with unilateral transtibial amputation reported inconsistent effects on the first peak vertical GRF when comparing DR feet to the SACH feet^{5, 20}.

The magnitudes of the fore-aft GRF peaks were significantly greater when the subjects walked with the flexion unit than when they walked without it (Table 4, Figure 4). The GRF is an indication of the accelerations of the body center of mass during walking²⁵, and reflects changes in mechanical energy of the inverted pendulum mechanism²⁶. The increased fore-aft GRF may indicate a more efficient gait (i.e., increased energy conservation) due to an improved foot rocker mechanism with the addition of the flexion unit. The effect of rocker radius on gait efficiency has been investigated in able-bodied subjects²⁷, but not in amputees. Nonetheless, it is presently not clear why the peak fore-aft forces were increased when the bilateral transtibial subjects in the current study walked with the flexion unit.

Decreased peak plantarflexion and dorsiflexion moments were observed when bilateral transtibial amputee subjects walked with the flexion unit compared to the baseline condition, though these results were not significant (Table 4, Figure 5). The ankle joint moment during stance phase is strongly influenced by the magnitude of the GRF and its position with respect to the joint center. Since the vertical GRFs were slightly increased when the subjects walked with the flexion unit, the reduction in ankle moments is attributed to reduced moment arms. During mid- to late stance phase, the flexion unit reduced the stiffness of the prosthetic ankle and prevented the center of pressure from progressing as far anteriorly under the foot as it did for the baseline condition. Consequently, the GRF vector passed closer to the ankle joint axis, reducing the ankle moment arm of the GRFs and producing a smaller ankle joint moment.

Joint power is equal to the joint moment multiplied by the relative angular velocity between two adjacent limb segments. It is important to note that the prosthetic foot and ankle are passive and cannot generate any power. Therefore, the ankle joint “power” of the prosthesis indicates the amount of the energy stored/dissipated and returned by the deflection of the prosthetic foot and ankle unit. The results indicate that peak ankle energy return (i.e. power generation) was greater when subjects walked with the flexion unit than when they walked without it (Table 4, Figure 6). The elastic components of the flexion unit increased the energy stored and returned at the ankle joint (Figure 6). When comparing similar studies involving persons with unilateral transtibial amputation, a majority reported a tendency to increase the magnitude of both positive and negative ankle joint power when subjects walked with DR feet compared to the SACH feet^{21, 28, 29}, although not all studies showed statistical significance. Whether the increased energy storage and return is beneficial for walking is currently not known. It is also unclear why the amputee subjects displayed greater peak hip power when they walked with the combined prosthetic configuration than the baseline configuration (Figure 7), since presumably an increase in energy return at the ankle joint would enable subjects to rely less on their hip musculature for producing energy for walking.

On the questionnaire, subjects reported that walking with the flexion unit felt smoother when walking with the flexion unit compared to the baseline condition. This perception may have been due to the restoration of the ankle rocker mechanism that serves to facilitate forward progression of the body over the supporting limb during single support phase. It may also relate to increased energy return by the flexion unit during late stance phase, which may have reduced the metabolic energy expenditure during ambulation and enabled the subjects to walk with less perceived effort and attain longer distances. The subjects also reported that the flexion unit made it easier for them to walk up/down stairs and inclines. The flexion unit permitted greater ankle sagittal plane motions required by those tasks. Gait analysis of the subjects walking up and down stairs and inclines would be useful to further evaluate the effects of increased prosthetic ankle plantarflexion/dorsiflexion.

The subjects also reported that, compared to the baseline configuration, the torsion unit reduced the perceived effort required to walk. The torsion unit may have improved more complicated walking tasks, such as turning and ambulating on uneven ground. One subject mentioned that the torsion unit was extremely beneficial when golfing. Presumably, those activities generate much greater torques in the transverse plane at the ankle than walking on level ground²⁴. Further investigation is required to better understand the perceived benefits provided by rotator units to lower-limb amputees.

Subjects indicated on the questionnaire that it felt easier to walk on uneven ground with the flexion, torsion, and combined configurations compared to the baseline configuration. The increased ankle motion with the flexion and torsion units may have allowed the prosthetic foot to better accommodate and maintain more uniform contact with uneven ground, and consequently may have decreased the moments acting about the prosthetic ankle and anatomical knee joints. It is important to recognize, however, that these perceptions may have been highly influenced by experiences outside of the research laboratory, thus possibly explaining why the quantitative data appear not to support subjective assessments. Gait analyses with subjects walking on uneven ground may prove to be quite valuable for evaluating the effects of prosthetic flexion and torsion units.

None of the subjects in this study preferred the baseline configuration over any of the other three. When comparing the flexion and the torsion configurations, about half of the subjects preferred the torsion configuration and the other half preferred the flexion configuration. Fourteen of 19 subjects liked the combined configuration best. It appeared that the flexion and torsion units both were perceived as being beneficial in providing comfort and improvements to the subject's general walking performance, and they felt most comfortable while walking with the combination of the two. Torburn et al.³ reported that all subjects preferred the prosthetic foot that enabled them to walk at the fastest speed, even if the change in speed was not significant. Similarly, most of the subjects in this current study expressed preference for the prosthetic condition that enabled them to walk at faster freely-selected speeds.

The results from this study should encourage clinicians to provide transtibial amputees with prosthetic ankle and foot systems that increase motions in the sagittal and transverse planes. Nonetheless, clinicians should perform individual and appropriate assessments of their clients to insure that they capable of using a component that may cause them to sacrifice some degree of stability for potentially improved mobility. Understanding about this trade-off between stability and mobility is currently lacking. The stiffness requirements for the prosthetic foot and ankle may very well be different for standing, for walking on level ground, and for ascending and descending slopes and stairs. Because joint stiffness requirements are known to vary during gait in able-bodied persons³⁰, components that are intended to improve ambulation may need to continuously adapt their stiffness in order to meet the demands required for the particular activity in which the users are engaged.

This current study has several limitations. First, most of the subjects who participated were younger than 70 years old (Table 1), yet the majority of persons who have lower-limb amputations are elderly and have etiology related to PVD. Therefore, our results may not be generalizable to the elderly population, since for the majority of elderly prosthesis users stability may be a more important goal rather than improved mobility. Nonetheless, like the younger subjects, the older individuals who participated in our study appeared to also benefit from the prosthetic ankle units. Another limitation of our study is that the questionnaires to record subjective perceptions were not validated or piloted for usability/reliability. The data collected for this study supplemented the quantitative gait measures and were not intended to provide stand-alone information. However, a better developed and validated questionnaire may be useful in the clinic to document prosthesis users' perceptions regarding their prosthetic configurations and could ultimately be used by clinicians to justify prosthetic prescriptions. A final limitation of our approach involves the lack of blinding of the subjects to the configurations tested. Since the prosthetic components added in this study were difficult to hide, no attempt was made to blind the subjects to the different configurations. As a result, the subjective observations may have been biased toward the components that were added and tested.

CONCLUSION

Analyzing the effects of increased prosthetic ankle motions in persons with bilateral transtibial amputations is beneficial because the absence of sound limb compensations enables the advantages and disadvantages of different prosthetic components to be more easily identified. When bilateral transtibial amputee subjects walked with the Endolite Multiflex Ankle, they showed a significant increase in ankle plantarflexion angle during loading response phase and in ankle dorsiflexion during pre-swing. Additionally, peak ankle joint moments were reduced and peak ankle powers were increased. When subjects walked with the Otto Bock Torsion Adapter, they displayed a significantly increased range of motion in the transverse plane of the ankle. The quantitative results of the current study indicated that the increased prosthetic ankle motion improved some gait parameters of persons with bilateral transtibial amputations. Subjects perceived that the flexion unit was beneficial for walking long distances, up and down stairs and inclines, and on uneven ground, while the torsion unit was useful for turning and walking on uneven ground. Transtibial amputees appear to benefit from prosthetic components that provide sagittal and transverse plane rotations at the ankle. However, clinicians are encouraged to perform individual and appropriate assessments of their clients to insure that they are capable of using a component that may cause them to sacrifice some degree of stability for potentially improved mobility.

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Figure 1.

Plot showing the mean patterns of sagittal plane ankle joint angle when the group of subjects walked with the baseline and flexion prosthetic configurations. The shaded area represents the mean pattern $\pm$ one standard deviation of able-bodied persons. The vertical line represents toe-off.

Figure 2.

Plot showing the mean patterns of transverse plane ankle rotation angle when the group of subjects walked with the baseline and torsion prosthetic configurations.

Figure 3.

Plot showing the mean patterns of vertical GRF when the group of subjects walked with the baseline and combined prosthetic configurations.

Figure 4. Plot showing the mean patterns of fore-aft GRF when the group of subjects walked with the baseline and flexion prosthetic configurations.

Figure 5.

Plot showing the mean patterns of ankle flexion/extension moment when the group of subjects walked with the baseline and flexion prosthetic configurations.

Figure 6.

Plot showing the mean patterns of ankle power when the group of subjects walked with the baseline and flexion prosthetic configurations.

Figure 7. Plot showing the mean patterns of hip power when the group of subjects walked with the baseline and combined prosthetic configurations.

Table 1

Subject's general information

Subject	Age	Gender	Height(cm)	Mass(kg)	Etiology
1	22	F	168	53	T
2	23	M	193	99	T
3	47	F	167	65	T
4	52	M	172	87	T
5	63	M	175	89	T
6	67	M	177	92	T
7	50	M	176	76	T
8	30	M	170	51	T
9	31	M	168	97	T
10	43	F	165	78	T
11	51	M	172	62	P
12	83	M	169	76	P
13	78	M	172	93	P
14	58	M	168	68	P
15	50	M	175	80	P
16	76	F	163	61	P
17	60	M	186	91	P
18	59	M	174	93	P
19	61	F	159	60	P
Mean	52.8		171.9	77.4	
SD	17.6		7.8	15.5	

M = Male, F = Female, T=Trauma, P= peripheral vascular disease

Table 2

Temporal-spatial data

Temporal-spatial data

	Baseline (SD)	Flexion (SD)	Torsion (SD)	Combined (SD)	AB (SD)
Walking speed (m/s)	0.91 (0.27)	0.92 (0.29)	0.92 (0.28)	0.95 (0.28)	0.82 (0.19)
Step length (cm)	56.9 (11.9)	57.0 (12.9)	57.0 (12.4)	58.8 (12.4)	58.2 (8.7)
Cadence (step/min)	93.8 (13.7) ^A	95.0 (15.1) ^A	94.4 (13.9) ^A	95.5 (14.2) ^A	84.3 (10.7) ^{B, F, T, C}
Step width (cm)	18.8 (4.8) ^{F, T, C, A}	17.1 (4.2) ^B	17.8 (4.7) ^{B, A}	17.4 (4) ^B	12.2 (3.2) ^{B, T}
Stance time (% gait cycle)	64.5 (3.8)	64.7 (3.6)	64.6 (3.5)	64.3 (3.3)	65.2 (3.7)
Double support time (% gait cycle)	14.5 (3.7)	14.6 (3.7)	14.5 (3.6)	14.2 (3.3)	15.1 (2)

AB = Able-bodied. Double support time is calculated by averaging the two periods of double support times in a gait cycle. The superscript indicates that the values were statistically significant (p<0.05) compared to those from other prosthetic configurations or able-bodied gait.

^B baseline,

^F flexion,

^T torsion,

^C combined prosthetic configurations,

^A able-bodied.

Table 3

Peak-to-peak kinematic data

Kinematic data					
	Baseline (SD)	Flexion (SD)	Torsion (SD)	Combined (SD)	AB (SD)
Ankle plantarflexion/dorsiflexion in stance phase (degrees)	12.5 (3.1) ^{F, C, A}	18.7 (4.9) ^{B, T}	12.1 (3.9) ^{F, C, A}	19.0 (4.8) ^{B, T}	20.2 (3.5) ^{B, T}
Ankle transverse plane rotation (degrees)	2.9 (1.5) ^{T, C, A}	2.7 (1.1) ^{T, C, A}	4.8 (2.7) ^{B, F, A}	5.2 (2.6) ^{B, F, A}	17.4 (6) ^{B, F, T, C}
Knee flexion/extension in stance phase (degrees)	12.5 (6.5) ^A	12.3 (6.1) ^A	12.3 (6.6) ^A	11.9 (6.7) ^A	16.1 (4.6) ^{B, F, T, C}
Knee flexion/extension in a gait cycle (degrees)	64.9 (11.8)	64.9 (11.4)	66.1 (9.9)	65.7 (11.5)	61.7 (4.1)
Hip flexion/extension (degrees)	43.6 (6.9) ^A	42.3 (5.6) ^A	42.9 (5.5) ^A	43.6 (6.4) ^A	36.5 (2.1) ^{B, F, T, C}
Pelvic obliquity in coronal plane (degrees)	8.4 (2.8)	8.4 (3.6)	8.3 (3.1)	8.4 (4.1)	6.3 (2.1)

The superscript indicates that the values were statistically significant ($p < 0.05$) compared to those from other prosthetic configurations or able-bodied gait.

^B baseline,

^F flexion,

^T torsion,

^C combined prosthetic configurations,

^A able-bodied.

Table 4

Kinetic data

	Kinetic data				
	Baseline (SD)	Flexion (SD)	Torsion (SD)	Combined (SD)	AB (SD)
First peak vertical GRF (BW)	1.09 (0.06) ^C	1.11 (0.17)	1.11 (0.1)	1.12 (0.12) ^B	1.05 (0.03)
Peak fore-aft acceleration GRF (BW)	0.11 (0.05) ^{F, C}	0.13 (0.05) ^{B, T}	0.12 (0.06) ^{F, C}	0.13 (0.06) ^{B, T}	0.12 (0.02)
Peak fore-aft deceleration GRF (BW)	0.10 (0.04) ^{F, C}	0.12 (0.06) ^{B, T}	0.11 (0.05) ^{F, C}	0.14 (0.05) ^{B, T}	0.11 (0.03)
Peak ankle plantarflexion moment (Nm/kg)	1.13 (0.16) ^A	1.08 (0.14) ^A	1.12 (0.15) ^A	1.06 (0.17) ^A	1.28 (0.14) ^{B, F, T, C}
Peak ankle dorsiflexion moment (Nm/kg)	0.19 (0.14) ^A	0.17 (0.09) ^A	0.19 (0.07) ^A	0.15 (0.05) ^A	0.10 (0.02) ^{B, F, T, C}
Peak positive ankle power (watt/kg)	0.38 (0.18) ^{F, C, A}	0.56 (0.3) ^{B, T, A}	0.36 (0.23) ^{F, C, A}	0.57 (0.31) ^{B, T, A}	1.26 (0.38) ^{B, F, T, C}
Peak negative ankle power (watt/kg)	0.79 (0.34)	0.94 (0.38) ^T	0.70 (0.37) ^F	0.88 (0.4)	0.71 (0.13)
Peak positive hip power (watt/kg)	0.96 (0.45) ^A	1.04 (0.48) ^A	1.00 (0.48) ^A	1.03 (0.45) ^A	0.50 (0.25) ^{B, F, T, C}
Peak negative hip power (watt/kg)	0.41 (0.33) ^{C, A}	0.52 (0.29) ^A	0.49 (0.32) ^A	0.52 (0.33) ^B	0.23 (0.09) ^{B, F, T, C}

The superscript indicates that the values were statistically significant ($p < 0.05$) compared to those from other prosthetic configurations or able-bodied gait.

^B baseline,

^F flexion,

^T torsion,

^C combined prosthetic configurations,

Table 5

Questionnaire results

Questionnaire Statements	Ankle Configurations			
	Baseline	Flexion	Torsion	Combined
If I have pain in my residual limb, this configuration reduces it.	2.5	3	3	3
This configuration is comfortable (increases my comfort) during walking.	4	4	4	4
This configuration makes my prostheses hard (harder) to swing as I walk.	2	2	2	2
This configuration enables me to walk long (longer) distances.	3^C	4	3^C	$4^B, T$
This configuration increases the effort to walk.	3^T	2	2^B	2
I am able to walk fast (faster) with this configuration.	4	4	3	4
Walking feels smooth (smoother) with this configuration.	3^F	4^B	4	4
This configuration makes me feel like I am stepping into a hole.	2	2	2	2
This configuration reduces twisting between my socket and residual limb.	2.5	3	3.5	3
This configuration feels comfortable (increase my comfort) during standing.	4	4	4	4
This configuration makes me feel unstable during walking.	2	2	2	2
This configuration allows me to be as active (more active) as I want to be.	3	4	4	4
This configuration enables me to turn easily (easier).	3^T	4	4^B	4
It is easy (easier) for me to walk up stairs with this configuration.	3^C	4	3	4^B
It is easy (easier) to walk down stairs or step off a curb with this configuration.	3	4	3	4
It is easy (easier) for me to walk up an incline with this configuration.	2^C	4	3^C	$4^B, T$
It is easy (easier) for me to walk down an incline with this configuration.	3^C	4	3^C	$4^B, T$
This configuration makes it easy (easier) for me to walk on uneven ground.	$2^F, T, C$	4^B	4^B	4^B
This configuration provides too much motion.	2	2	2	2
This configuration doesn't provide enough motion.	$4^F, T, C$	2^B	2^B	2^B
This configuration makes me feel like I'm walking up hill.	2	2	2	2
This configuration makes me feel like I'm walking down hill.	2	2	2	2

Questionnaire Statements	Ankle Configurations			
	Baseline	Flexion	Torsion	Combined
This configuration makes me stub my toe (more) during swing.	2	2	2	2
Overall, this configuration provides me with sufficient (greater) comfort.	4	4	4	4

Median values for subject responses are provided. The superscript indicates that the values were statistically significant ($p < 0.008$) compared to those from the other prosthetic configurations.

Key: 1-“I strongly disagree”, 3-Neutral, 5-“I strongly agree”.

B_{baseline} ,

F_{flexion} ,

T_{torsion} ,

$C_{\text{combined prosthetic configurations}}$.

Table 6

Perceptions of subjects about the different interventions.

Baseline
<ul style="list-style-type: none"> • The forefoot of the prosthetic feet were rigid and difficult to roll over. • Subjects tired easily. • Subjects needed a wider base of support for stability when walking. • Subjects needed extra effort to climb stairs. • Subjects had better standing stability.
Flexion
<ul style="list-style-type: none"> • Several subjects experienced a sense that they might fall backwards during standing. • One subject felt more stable during standing and walking with this configuration. • Flexion configuration made walking smoother. • Subjects felt easier to climb up and down stairs and inclines.
Torsion
<ul style="list-style-type: none"> • It made walking and turning easier. • One subject needed more than two weeks to adjust to the feet. • There was too much rotational motion occurred at the wrong time, especially when standing and ascending stairs.
Combined
<ul style="list-style-type: none"> • The combined configuration allowed them to walk more naturally. • The configuration better replicated the motion they possessed prior to amputations. • One subject reported feeling too much motion at the ankle. • The sensation of falling backward continued to be a concern