

A preliminary investigation of pelvic obliquity patterns during gait in persons with transtibial and transfemoral amputation

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Abstract—Differences in pelvic obliquity between small groups of persons with unilateral lower limb amputation and subjects without amputation were analyzed. Kinematic walking data were collected as six males with transtibial amputation and three males with transfemoral amputation walked over a range of speeds. The pelvic obliquity patterns and amplitudes from the groups with amputation were compared to normal data. Results showed that smaller peak-to-peak amplitudes of pelvic obliquity were associated with higher amputation levels. Pelvic drop during early prosthetic-limb stance tended to be smaller than during early sound-limb stance. Most of the subjects with amputation exhibited an obliquity pattern in which the hip on the prosthetic side was raised above the stance-side hip during prosthetic swing phase, indicative of a compensatory action known as hip-hiking. The subjects with transfemoral amputation exhibited this hip-hiking pattern during sound-limb swing phase as well. Results from this study suggest that further investigation is required to determine those limitations of current prosthetic technology that adversely affect pelvic obliquity in the gait of persons with amputation, and to determine if significant benefit can be realized by restoring a normal pattern of pelvic obliquity to the gait of persons with amputation.

Key words: *gait, pelvic obliquity, prosthetics, transfemoral amputation, transtibial amputation.*

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INTRODUCTION

Gait analysis investigations have not established normative data for the walking patterns of persons with transtibial or transfemoral prostheses. Prosthesis variations and individual differences may preclude the identification of normative patterns. Previous investigations of the gait of persons with amputation have usually focused on sagittal plane movements. In this paper, we examined possible trends in coronal plane movement, specifically pelvic obliquity, for a small group of persons with amputation. We believe general trends in coronal plane movement may help explain some aspects of their walking and provide insight as to how to improve prosthetic gait.

Pelvic obliquity is the coronal-plane rotation of the pelvis, defined as the angle between the horizontal plane and the medial-lateral axis of the pelvis. In normal gait, the pelvic obliquity curve is periodic with one cycle per stride (**Figure 1**). As the heel of the leading limb contacts the ground, the pelvis is nearly neutral. During loading response, the hip of the trailing leg begins to drop, increasing the magnitude of pelvic obliquity. Immediately after toe-off, pelvic obliquity reaches its greatest amplitude. The motion of the pelvis is then reversed, with the swing-side hip regaining its neutral position during midstance, and then lowering slightly just before heel contact. The peak of the “primary” component of pelvic obliquity occurs just after toe-off, while smaller fluctuations (“secondary” components), occur near the end of the single-support phase. Subjects without amputation walking at freely-selected speeds typically exhibit 5° to 7° of pelvic obliquity toward each side (1–3), and the magnitude of pelvic

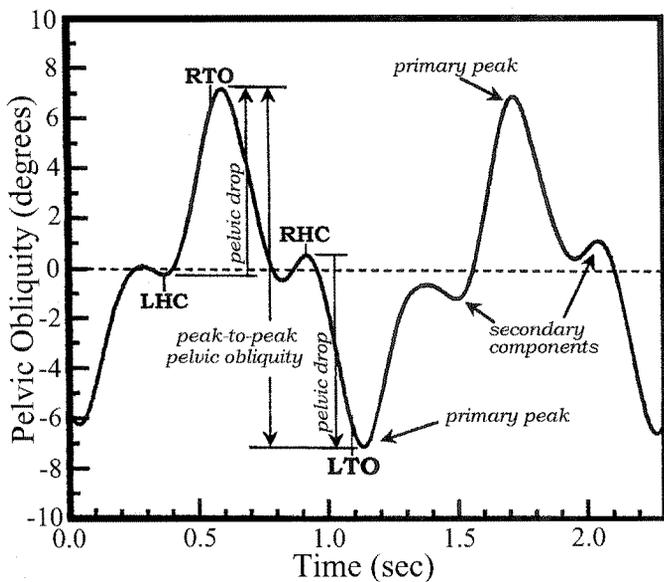


Figure 1.

Pelvic obliquity recorded from two gait cycles of a healthy male subject walking at 1.4 m/s (3). The major events of the gait cycle are labeled as: R=right, L=left, HC=heel contact, TO=toe-off. The global maxima and minima are termed the "primary" components, while the smaller fluctuations are termed the "secondary" components. The measurements of pelvic drop and peak-to-peak pelvic obliquity are illustrated.

obliquity has been observed to increase linearly with the walking speed (3).

Some gait researchers have suggested that pelvic obliquity serves a shock-absorbing function during the loading response of normal walking (2-4). As weight is transferred from the trailing to the leading leg, the hip abductor musculature of the lead leg restrains the drop of the head, arms, and trunk (HAT). This mechanism decreases the shock forces experienced by the body in normal walking. Proponents of the six determinants of gait (1) claim that the purpose of pelvic obliquity is to decrease the peak-to-peak amplitude of the body's center of mass during normal gait, but recent work questions this concept (3). The timing of the pelvic obliquity rotation within the normal gait cycle prohibits it from decreasing the amplitude of the trunk's vertical movement.

Since the weight of the upper body acts downward through the pelvis, control of pelvic motion is vital to maintaining whole body balance in the coronal plane (5). Pelvic control in this plane may be a significant concern for those with compromised muscular and skeletal structure, as in, for example, persons with lower limb amputation. Issues of motion and stability in the coronal plane have been virtually ignored in the literature on the gait of persons

with amputation.

This article reports results from a preliminary investigation into the nature of pelvic obliquity among persons with unilateral transtibial and transfemoral amputation. The study was designed to assess the qualitative and quantitative differences in pelvic obliquity between small groups of subjects with transtibial amputation, transfemoral amputation, and no amputation. It is hypothesized that the pattern of pelvic obliquity is disrupted in the gait of those with amputation, and that restoration of this shock-absorbing mechanism may be important for obtaining a more functional and more aesthetic gait.

METHODS

Subjects

Nine adult males with unilateral amputation (six transtibial and three transfemoral) participated in the study (Table 1). Four were students at the School of Prosthetics and Orthotics at Northwestern University, and the other five were patients of the Prosthetics-Orthotics Clinical Services at the Rehabilitation Institute of Chicago. The subjects were selected based on their good health and high activity level, ability to walk at a variety of speeds without walking aids, and having a lower limb amputation with non-vascular related etiology. All of the subjects signed consent forms prior to participating in the study.

Data Collection and Processing

Each subject was asked to come to the Human Mechanics Measurement Laboratory (HMML) located in the Northwestern University Prosthetics Research Laboratory and Rehabilitation Engineering Research Center (NUPRL & RERP) for one data collection session. The HMML is equipped with a 10-meter conductive walkway with two embedded AMTI force platforms. A CODA 3 Motion Measurement System (6), which tracks the positions of retro-reflective prismatic markers, was positioned at one end of the walkway.

The experimental protocol for this investigation is similar to that previously described for a study of pelvic obliquity in healthy ambulators (3). Two markers were attached to each subject via a Velcro® belt wrapped securely about the subject's pelvic girdle; the markers were placed over the right and left anterior superior iliac spines (ASIS). The positions of the two ASIS markers were averaged in post-processing of the data to define the position of a virtual marker, called the 'pelvic origin'. During the data collection, the subjects walked on their own prostheses and wore shorts.

Table 1.
Subject data.

Subject	Amputation Level/Side	Age (yrs)	Height (m)	Weight (kg)	Residual Limb Length (m)	Socket Type	Foot Type	Knee Type
1	TT/left	26	1.88	81.5	0.191	PTB-suction	VSP flex-foot	—
2	TT/right	27	1.68	81.6	0.108	PGB-sleeve	greissinger	—
3	TT/left	32	1.65	61.7	0.130	PTB-SC	airflex	—
4	TT/right	45	1.83	86.8	0.178	PTB-SC	flex-foot	—
5	TT/right	54	1.87	90.4	0.102	PTB-SC	seattle-foot	—
6	TT/right	65	1.78	92.0	0.125	PTB-SC	carbon copy II	—
7	TF/left	40	1.82	73.6	0.343	IC	flex-foot	hydraulic
8	TF/left	44	1.78	74.9	0.415	IC	flex-foot	hydraulic
9	TF/right	73	1.70	70.8	0.370	QUAD	settle-foot	hydraulic

TT=transtibial; TF=transfemoral; PTB=patellar tendon bearing; IC=ischial-containment; QUAD=quadrilateral; suction, sleeve, and SC (supra-cond-ylar) indicate suspension type.

Most of the subjects with amputation wore athletic shoes, but two chose to wear their everyday hard-soled dress shoes in order to avoid a need for prosthetic alignment changes.

Each subject was asked to walk at four speeds: first at his freely selected (termed “normal”) walking speed, then at “slow” (approximately 20 percent slower than freely selected), “fast” (approximately 20 percent faster) and “maximum” (fastest speed attainable without running) walking speeds. Subjects performed at least 12 walking trials at each speed. One of the investigators used a stopwatch to monitor walking speed by timing the subjects as they traversed a known distance along the walkway; the subject was given verbal cues to help him maintain a near-constant speed (± 5 percent of target speed) between the trials. For analyzing the data, walking speed was also calculated as the mean value of the subject’s instantaneous forward velocity during steady state gait. This velocity was obtained by taking the first derivative of the pelvic origin marker coordinate data in the direction of forward progression.

Three-dimensional kinematic data were collected as the subject walked along the walkway toward the CODA. The marker positions were sampled at 300 Hz; the precision of the measured marker position was ± 1 mm. The raw position coordinate data were bidirectionally filtered using an effective tenth-order Butterworth low-pass filter with a cut-off frequency of 6.0 Hz and having zero phase shift.

Calculation of Pelvic Obliquity

Pelvic obliquity was calculated as the angle in the coronal plane of the walking subject that is formed between the line connecting the two ASIS markers and a horizontal plane. Pelvic obliquity (θ) was calculated by:

$$\theta = \frac{180}{\pi} \cdot \arctan \frac{(Y_S - Y_P)}{\sqrt{(X_S - X_P)^2 + (Z_S - Z_P)^2}} \quad [1]$$

where (X_S, Y_S, Z_S) and (X_P, Y_P, Z_P) are the three-dimensional (3D) coordinates of the sound and prosthetic-side ASIS markers, respectively (**Figure 2**). When the ASIS markers were at the same elevation, pelvic obliquity was 0° ; when the prosthetic-side ASIS marker dropped below the sound-side ASIS marker, pelvic obliquity was considered positive. For unimpaired subjects, the sign of pelvic obliquity was considered positive when the right ASIS marker was below the left.

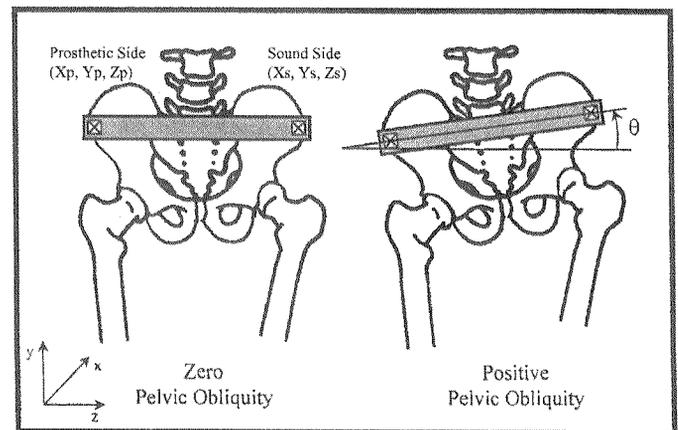


Figure 2.

Pelvic obliquity is calculated from the three-dimensional coordinates of the prosthetic- and sound-side anterior superior iliac spine (ASIS) markers. The coordinate system of the laboratory is shown, where the x, y, and z axes denote the anterior-posterior, vertical, and medial-lateral directions, respectively.

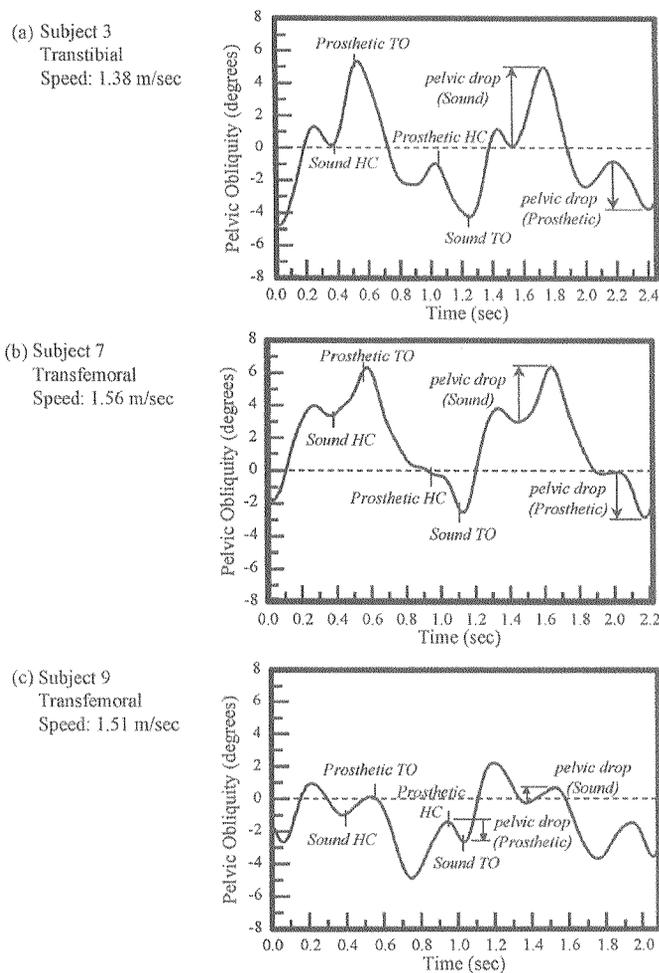


Figure 3.

(a) Pelvic obliquity recorded from two gait cycles of subject 3 (TTA). Five subjects with TTA exhibited this type of pattern. (b) Two gait cycles of pelvic obliquity recorded from subject 7 (TFA). Subject 8 exhibited similar pelvic obliquity characteristics. Note specifically the asymmetry in the pelvic drop between the prosthetic and sound loading response phases in (a) and (b). (c) Pelvic obliquity recorded from two gait cycles of subject 9 (TFA). Note the relatively small peak-to-peak amplitude of the pelvic obliquity curve as well as the diminished pelvic drop values. In fact, the peak values of pelvic obliquity occurring in the middle of stance phase actually exceeded those occurring just after toe-off. Subject 9 was the only subject with TFA to wear a quadrilateral socket.

Data Analysis

For each walking trial, the 3D positions of the markers were recorded along the entire length of the walkway. A period of steady-state walking, which encompassed at least two gait cycles per trial, was defined as a period in which forward walking speed remained relatively constant; pelvic obliquity was examined only during this interval.

Kinematic data from the gait of three, unimpaired, adult male subjects were previously collected in our laboratory (3) using an experimental protocol similar to the one

employed in the present study. This normative data was used as a reference for speed-based comparisons of the amputee gait data.

Qualitative differences in the pelvic obliquity patterns between the transtibial, transfemoral, and unimpaired subject groups were noted. Pelvic drop (2) was defined as the angle through which the contralateral hip rotated downward from the time of ipsilateral foot contact to maximum obliquity on that side (see **Figure 1**). Pelvic drop was calculated during both the sound and prosthetic stance phases of each gait cycle, and the values were averaged over the duration of each trial's steady-state period. Paired t-tests were performed on each subject's data to determine if there was a significant difference ($p=0.05$) between values of pelvic drop during initial sound-side stance versus initial prosthetic stance.

The differences between the successive primary maxima and minima of the pelvic obliquity curve were calculated during the steady-state period and averaged to obtain a mean value of peak-to-peak pelvic obliquity per walking trial. These peak-to-peak amplitudes were plotted against walking speed for each group of subjects, and then regression equations were fitted to the data in order to make comparisons of peak-to-peak obliquity over the range of walking speeds.

RESULTS

The pelvic obliquity waveforms of the subjects with amputation were different from those of the unimpaired subjects. Specifically, the pelvic obliquity curves recorded from the subjects with transtibial amputation (TTA) were asymmetric about neutral. In all of the subjects with TTA, except subject 6, pelvic drop during the prosthetic loading response was significantly less than that which occurred during the sound-side loading response across the entire range of walking speeds. This is illustrated in **Figure 3a**, which shows a response typical of subjects 1–5. The obliquity patterns in five of the six subjects with TTA indicated mild hip-hiking during prosthetic swing phase; that is, the swing-side (prosthetic) hip was raised slightly above the stance-side (sound) hip before prosthetic heel contact (note dotted lines in **Figure 3a**). Subject 1 (TTA) did not display this hip-hiking pattern.

Subject 1 exhibited a symmetric obliquity curve, but with a relatively large steady-state offset of approximately $+7^\circ$ in the baseline. The positive sign of the offset indicated that the prosthetic-side hip was consistently lower than the sound-side hip. Static pelvic obliquity was measured in subjects with and without amputation during quiet standing.

Non-zero values of the static pelvic obliquity were assumed to be due to imperfect placement of the ASIS markers and/or leg-length discrepancy. The amplitude of a subject's static pelvic obliquity could usually be correlated with the magnitude of his pelvic obliquity offset that was measured during the walking trials. However, the static pelvic obliquity measured in subject 1 (approximately 2°) did not account for the large steady-state offset (approximately 7°) observed in his walking data. Interestingly, subject 1 was the only subject with TTA in this study who walked with a vertical shock-absorbing pylon (Flex Foot Re-Flex Vertical Shock Pylon), which is a prosthetic component that serves to decrease shock through pylon shortening.

The pelvic obliquity curves measured in the subjects with transfemoral amputation (TFA)—that is, subjects 7, 8, and 9—deviated substantially from those of the unimpaired subjects. A pelvic obliquity curve typical of those recorded from two of the subjects with TFA (subjects 7 and 8) is shown in **Figure 3b**. These subjects demonstrated a pattern of hip-hiking during single support on both the sound and prosthetic limbs in which the swing-side hip was consistently elevated above the support-side hip throughout the swing phase (note the dotted lines in **Figure 3b**). They also exhibited a smaller amount of pelvic drop during the loading response on the prosthetic limb than occurred during loading of the sound limb. This difference in the amount of pelvic drop between the sound and prosthetic sides during the loading response was noted across the entire range of walking speeds tested; it was statistically significant in subject 7, but not for subject 8. The values of pelvic drop calculated for the subjects with TFA were consistently smaller than those measured from the unimpaired subjects.

The difference between the pelvic obliquity curves of subject 9 and those of the other two subjects with TFA lies in the amplitude and timing of the peak values of pelvic obliquity (**Figure 3c**). In the pelvic obliquity waveforms of subject 9, the amplitude of the secondary peaks occurring during mid- to terminal stance actually exceeded the amplitudes of the primary ones occurring just after toe-off (**Figure 3c**); this result was consistent across his full range of walking speeds. This subject also exhibited greater pelvic drop during the loading response on the prosthetic limb than during the corresponding phase on the sound limb ($p < 0.01$). However, pelvic drop values on the prosthetic and sound sides ranged from 1° – 3° . Although it wasn't particularly evident through a visual observation of the subject's gait, the measured pelvic obliquity waveform for this subject indicated hip-hiking compensation on both

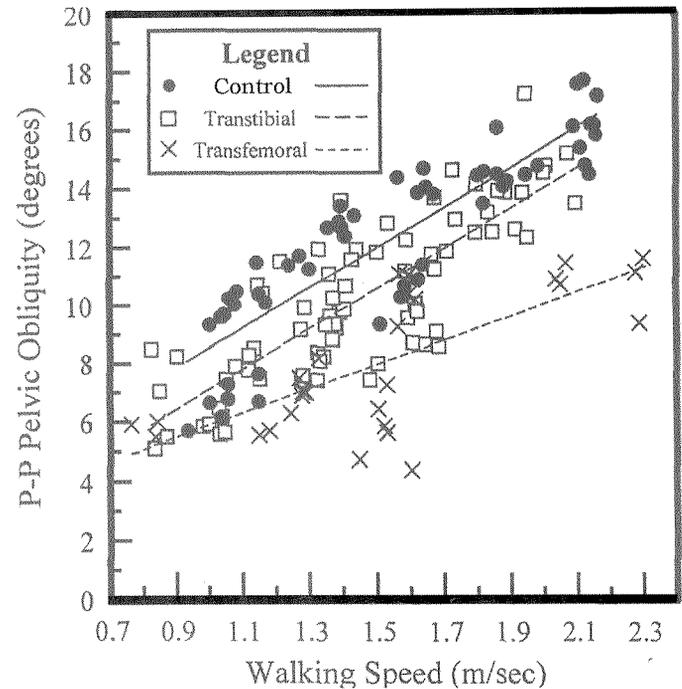


Figure 4.

Peak-to-peak pelvic obliquity versus walking speed for the subjects with TTA, the subjects with TFA, and subjects with no amputation. Linear regression analysis on the data sets revealed similar slopes between the data of subjects with no amputation and those with TTA, and similar intercepts between the data of subjects with no amputation and those with TFA.

sides of the body. The swing-side hip was not substantially lowered just after toe-off, as confirmed by the very low pelvic drop values, and it was raised above the stance-side hip during the middle of single limb stance. This mechanism was presumably used to either increase toe clearance of the swing leg, or it could be evidence of a lateral trunk lean employed by the subject to increase stability over the prosthesis.

A plot of the peak-to-peak pelvic obliquity versus walking speed for the unimpaired subjects and those with TTA and TFA is shown in **Figure 4**. The data of the subjects with amputation exhibited greater inter-subject variability than the normal data. The regression lines in **Figure 4** indicate a linear relationship between the peak-to-peak pelvic obliquity and the walking speed (correlation coefficients all exceed 0.70). The regression line for the subjects with TTA has a slope similar to the regression line for the unimpaired subjects, but has a significantly lower intercept. Conversely, the regression line for the subjects with TFA has an intercept similar to that for the unimpaired subjects, but has a significantly lower slope. Most of the

subjects with TFA had smaller peak-to-peak pelvic obliquity amplitudes than either the subjects with TTA or the unimpaired subjects at walking speeds above 1.0 m/s. In fact, even at the fastest speeds of the subjects with TFA, the largest values of peak-to-peak pelvic obliquity were less than 12°.

DISCUSSION

The locomotor mechanism is altered by lower limb amputation, and the alteration tends to increase with the level of amputation. In persons with amputation, it has been shown that there are decreases in both the freely selected and maximum walking speeds with increasing levels of amputation (7–15). Persons with amputation appear to naturally select a speed of walking in which the rate of energy expenditure is similar to that of persons without amputation (8,16). However, the energy expenditure per unit distance walked (the energy cost with respect to walking speed) progressively increases with higher amputation levels (8,16,17). While the exact cause for the increased energy expenditure in persons with amputation is still uncertain, it does not appear to be related to an increased vertical excursion of the body center of mass, as some researchers have suggested (18). Part of the reason why they expend greater energy during gait and walk more slowly than unimpaired persons may be attributable to the altered dynamics of their locomotor mechanism that result from constraints imposed by their leg prostheses. Specifically, an inability to rotate the pelvis in the frontal plane, reflected by an altered pelvic obliquity waveform, may reduce the shock absorption capability of the locomotor mechanism at heel contact and during the loading response.

Pelvic Obliquity during Loading Response

In all of the data pertaining to the subjects with amputation, there was a distinct difference in the magnitude of pelvic drop that occurred during the loading response of the sound limb compared with that of the prosthetic limb. In five of the subjects with TTA and one with TFA, the magnitude of pelvic obliquity from the time of ipsilateral heel contact to the maximum value just after contralateral toe-off was significantly smaller for the prosthetic limb than for the sound limb.

All the pelvic drop values for the subjects with TFA were much less than those measured from the subjects without amputation. During both periods of loading response, the pelvis typically rotated only a few degrees in the coronal plane. Two possible explanations for these

decreased pelvic drop values are:

1. During the prosthetic loading response, the proximal rim of the transfemoral socket may have made prosthetic-side hip adduction uncomfortable for the user, so he didn't allow his pelvis to drop.
2. During the sound-limb loading response, which corresponds with the prosthetic pre-swing phase, the subjects with TFA may have been unable to achieve prosthetic knee flexion sufficient to permit much pelvic drop.

An overly extended prosthetic knee in late stance caused by difficulty with initiating prosthetic knee-break could delay the onset of swing-phase leg-shortening and prevent the pelvis from dropping as much as in normal ambulators.

Pelvic Obliquity during Single-Limb Stance

Pelvic obliquity in the subjects with TTA tended to be abnormal during the middle of sound stance because the swing-side hip was raised above its stance-side counterpart (**Figure 5b**), a compensatory action known as hip-hiking. It may be that, due to the inability to dorsiflex the prosthetic ankle, the subjects with TTA lifted their prosthetic limbs slightly with their pelvis during swing phase to insure adequate foot clearance (2,19).

The pelvic obliquity of the subjects with TFA tended to deviate from normal during both sound *and* prosthetic single-limb stance. During each single-limb stance phase, the swing-side hip was raised above the elevation of the stance leg hip (**Figure 5c**). This distortion of the pelvic obliquity plot could be due to an active hip-hiking compensation (i.e., active use of stance-side hip abductor musculature), a passive leaning or lateral flexion of the trunk, or some combination thereof. The use of a leaning or lateral trunk flexion mechanism during prosthetic stance would reduce the coronal-plane hip moment and avoid reliance upon the abductor musculature of the amputated side for stability.

Peak-to-Peak Pelvic Obliquity

There was a consistent difference in the peak-to-peak magnitudes of pelvic obliquity between the gait of the subjects with and without amputation throughout the range of walking speeds considered. The peak-to-peak pelvic obliquity values of the subjects with TTA were less than or equal to those of the three unimpaired subjects, but the values of the subjects with TFA were consistently less than the normal range. Thus, higher levels of amputation appear to be associated with smaller amplitudes of pelvic obliquity. We believe the decreased pelvic obliquity is due in part to the compensatory actions (hip-hiking) utilized to increase

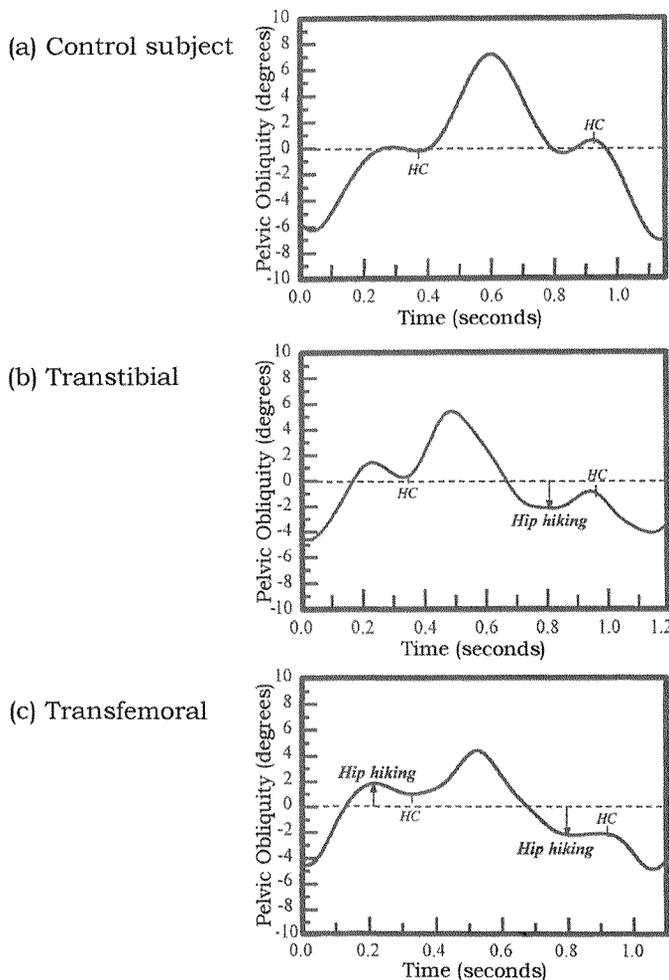


Figure 5. One cycle of pelvic obliquity for (a) one unimpaired subject, (b) subject 3—TTA, and (c) subject 7—TFA. Time of heel contact (HC) is identified. The dashed line represents a “neutral” orientation of the pelvis. Just prior to each heel contact, the pelvis of the subject without amputation regains, or slightly exceeds, its neutral orientation. The pelvic obliquity data from the subjects with amputation indicated that subjects with TTA utilized hip-hiking during prosthetic swing phase, while the subjects with TFA used hip-hiking during the swing phase of both the sound and prosthetic limbs.

prosthetic foot clearance during swing. Hip-hiking requires that the ambulator work against gravity and forcibly change the natural dynamics of the locomotor mechanism in order to raise the hip of their swing leg. By preventing a normal amount of pelvic drop during the loading response and utilizing hip-hiking during the swing phase, the primary components that appear in the normal pelvic obliquity waveform are diminished while the normally small secondary components become dominant and ultimately determine the peak-to-peak magnitude of the waveform.

Pelvic Obliquity and Shock Absorption

Most of the subjects with amputation in the present study were unable to attain walking speeds as high as the subjects without amputation. The results have also demonstrated that the former group exhibited distinct differences in pattern and magnitude of pelvic obliquity compared with the latter. Since one of the primary functions of the lower limb is to absorb impact forces at heel contact (2,20-22), the gaits of some persons with amputation may be compromised due to the loss of physiological shock-absorbing mechanisms, specifically a normal pattern of pelvic obliquity.

Cappozzo (23) observed that subjects with TFA walking at their maximum speed exhibited vertical accelerations with magnitudes similar to those of unimpaired subjects walking at their respective maximum speeds. Cappozzo speculated that the vertical acceleration of the body, and the mechanical load associated with it, might be one of the limits to walking speed. That is, persons with lower limb amputation may walk more slowly to reduce the impact and jarring sensations to the body that they feel with each step on the prosthesis. Wirta et al. (24), conducted a study in which persons with TTA walked with various foot-ankle devices and found that they preferred walking with the mechanism that produced the least shock and had the greatest damping. It could be that the manifestations of gait abnormalities observed in this study represent the subject’s solution to the necessary compromise that must occur between comfort and economy. If pelvic obliquity serves to absorb shock during the loading response phase of gait, as researchers have suggested (2-4), then perhaps persons with lower limb amputation walk more slowly to accommodate for the lack of shock absorption normally provided by the pelvic obliquity mechanism. The control of shock levels may be more important than the control of energy levels.

Pelvic Obliquity and Vertical Shock-Absorbing Pylons

In subject number 1 (TTA), the pelvic obliquity offset during walking seemed too large to be explained merely by marker placement error. His data indicated that the prosthetic-side hip was consistently lower than the sound-side hip throughout the gait cycle. Subject 1 was the only subject with TTA to walk with a vertical shock-absorbing pylon. The Flex Foot Re-Flex Vertical Shock Pylon (VSP), like other vertical shock-absorbing pylon designs currently on the market, is a telescoping mechanism that shortens under applied load, decreasing the overall length of the prosthesis for the entire duration of prosthetic stance phase. During the prosthetic stance phase of gait, the compression

of the vertical shock-absorbing pylon would contribute to a decrease in the functional prosthetic limb length, creating a dynamic leg-length discrepancy. A persistent offset in the pelvic obliquity during gait has been previously observed when there is a moderate shortening in one of the legs (25,26). Because of the functional leg length discrepancy of the prosthetic limb, the person may have problems with sound-limb foot clearance. Some manufacturers of vertical shock-absorbing pylons instruct prosthetists to fabricate the prosthesis so that it is slightly longer than the sound limb to avoid problems with sound-limb foot clearance when the pylon compresses during prosthetic stance phase. However, if the prosthetic length is increased too much, the length of the prosthesis may be too long during swing phase, forcing the wearer to compensate for prosthetic foot clearance. Since several adverse long-term effects of walking with an effective leg-length discrepancy have been reported in the literature (27), further investigation into the effect of added vertical compliance on the walking mechanism of the person with unilateral amputation seems warranted.

Pelvic Obliquity and the Quadrilateral Socket

At all of the walking speeds for subject 9 (TFA), the pelvic obliquity was greatest during the middle of the stance phase; the magnitude and sign of pelvic obliquity at that point indicated that the swing-side hip was raised several degrees above the stance-side hip. Much smaller maxima and minima occurred just after toe-off, which is normally when the maximum pelvic obliquity is observed in subjects without amputation (3,4). Subject 9 was the only subject with TFA who exhibited this altered pattern of pelvic obliquity throughout his range of walking speeds, and he had the least peak-to-peak pelvic obliquity of all our subjects. Coincidentally, he was fitted with a quadrilateral socket, while the other two subjects with TFA were fitted with ischial-containment sockets. The quadrilateral socket is designed to place the ischium on the posterior-medial brim of the socket during stance phase, although there was no validation of this occurrence in the present study. The biomechanical constraints imposed by the quadrilateral socket may make it difficult to achieve a normal pattern of pelvic obliquity. Because of its relatively large mediolateral dimension, the quadrilateral socket has been criticized for being ineffective at controlling stability in the coronal plane (28–30). Studies of persons with TFA walking with ischial-containment and quadrilateral sockets have shown that quadrilateral sockets introduce or exacerbate gait deviations and significantly increase the energy expenditure of gait (30,31). Further investigation is required to determine the

influence of prosthetic socket design on pelvic obliquity during the gait of persons with TFA.

Pelvic Obliquity and Prosthetic Foot Type

The subjects with TTA who participated in this study all used different prosthetic feet. Some studies have shown that prosthetic foot choice can significantly affect gait characteristics among persons with amputation (32–36). However, other researchers (37) found “no clinically significant changes in gait” in a group of persons with TTA wearing five different prosthetic feet. Prosthetic foot type may have great impact on the gait of persons who use lower limb prostheses. More studies of prosthetic gait that control for foot type are needed.

CONCLUSIONS

Summary of Significant Findings

This study characterizes pelvic obliquity patterns in a small group of subjects with unilateral transtibial and transfemoral amputation. The findings considered most significant include:

1. Pelvic obliquity waveforms in both groups with amputation were different from normal patterns, and were characterized by decreases in the peak-to-peak amplitudes of pelvic obliquity with higher levels of amputation.
2. Pelvic drop tended to be less during early prosthetic-limb stance than during early sound-limb stance in both the subjects with TTA and with TFA.
3. The pattern of pelvic obliquity observed in one subject with TTA using a vertical shock-absorbing pylon suggests that further investigations are needed to determine the effect of these telescoping mechanisms on the gaits of subjects with unilateral or bilateral amputation.
4. A deviant pattern of pelvic obliquity and its temporal relationship with the gait cycle observed in the one subject with TFA wearing a quadrilateral socket suggests that this socket design may impede coronal-plane pelvic movement.
5. It is hypothesized that the slower speeds of walking typically exhibited by persons walking on prostheses are related to decreased shock absorption qualities of the gait.

Implications

A primary goal of prosthetic fitting and alignment is to reestablish a "normal" gait pattern for the person with amputation. A more aesthetically pleasing gait may naturally result if functional considerations are addressed first (38). Establishing a normal pattern and magnitude of pelvic obliquity may be an effective means to help accomplish this goal. The restoration of physiological shock absorption mechanisms in persons with amputation is an important goal, because it decreases the potential for impact injury to the intact joints and the soft tissues of the residual limb. Restoring the shock absorption capabilities of persons with amputation may increase their freely selected speed of walking, as well as raise the upper limit of their range of comfortable walking speeds. To restore a normal pattern of pelvic obliquity, the issues of socket design and prosthetic knee/ankle design must be addressed. First, the proximal portion of a socket should allow the contralateral hip to drop during early prosthetic stance. Secondly, to allow the hip on the prosthetic side to drop, the prosthetic knee and ankle should allow late-stance knee flexion while the limb is still partially loaded. Creating a prosthesis that would allow a person with amputation to walk with a normal pelvic obliquity pattern could be an important step in the improvement of that person's gait.

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