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Step by Step: A Study of Step Length in Able-bodied Persons, Race walkers, and Persons with Amputation

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ABSTRACT

Step by Step: A Study of Step Length in Able-bodied Persons, Race walkers, and Persons with Amputation

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Step length is a common measurement taken during gait analyses. It allows one to determine asymmetries between the two legs, compare differences between subjects, and even compare intra-subject differences for changing parameters. Yet there has been little investigation of step length specifically and how it is modulated during walking. This dissertation explores the methods by which different groups of people modulate their step length. It was hypothesized that step length is modulated by several different means: increasing hip flexion and extension, increasing ankle-foot roll over arc length; increasing stance foot heel rise to further extend the trailing limb, and increasing pelvic rotation.

Gait analyses were performed for able-bodied persons, race walkers, persons with bilateral trans-tibial amputation and persons with partial foot amputation. Along with temporospatial and kinematic data, ankle-foot roll over shapes and Segment Contributions to Step Length (SCSL)

were determined. The SCSL analysis was introduced as a method to examine how each of the lower limb segments contributes to the overall step length and how these contributions vary for different walking conditions (e.g. different speeds or prosthetic devices). The SCSL analysis was also used to compare the differences in segment contributions between subject groups.

Results found that for a range of step lengths, percent contribution of the lower limb segments was fairly constant for able-bodied walking. Persons with normal effective foot lengths (e.g. intact feet) are able to utilize the ankle-foot segment to modulate step length, while those with shorter effective foot lengths displayed higher percent contributions from other lower limb segments. Largest contributions for all subjects were from the shank and thigh segments, though differences in contribution by the trailing and leading limbs were observed between subject groups. Although an increase in pelvic rotation contributed to an increased step, it appears to play a smaller role than previous studies seem to suggest.

The SCSL analysis is a simple tool to analyze step length contributions. By knowing the differences in segment contributions of persons with gait pathology, we can better determine what treatments or training procedures can improve upon gait.

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List of Abbreviations

AB	Able-bodied
AFO	Ankle foot orthosis
COP	Center of pressure
BTT	Bilateral trans-tibial
BTTA	Bilateral trans-tibial amputation/amputee
EFL	Effective foot length
EFLR	Effective foot length ratio
GC	Gait cycle
GRF	Ground reaction force
LL	Leg length
PFA	Partial foot amputation/amputee
SCSL	Segment contribution to step length
SL	Step length
vGRF	Vertical Ground Reaction Force

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Chapter 1: Introduction

Don't be afraid to take a big step if one is indicated. You can't cross a chasm in two small jumps.

-- David Lloyd George

Step length, the distance traversed during gait for one step, is a common measurement taken during gait analyses. It allows one to determine asymmetries between the two legs, compare differences between subjects, and even compare intra-subject differences for changing parameters such as increasing walking speed or wearing different types of shoes. Yet there has been little investigation of step length specifically and how it is modulated during walking. Changes such as an increase in step length can be beneficial for many different types of subjects: from persons with unilateral lower limb amputations who may walk asymmetrically between their sound and prosthetic side, to persons with bilateral amputations who tend to display shorter step lengths than able-bodied ambulators. Even able-bodied subjects may wish to walk farther or faster over a given time.

		Average speed (m/sec)	Average step length (m)	% Stance Phase	Difference in % stance between sides
1	Able-bodied (9 subjects, internal data)	1.41	0.76	62 %	0 %
2	Able-bodied (James and Oberg 1973)	1.51	0.78	61%	0 %
3	Bilateral (below-knee) (Su 2004)	1.14	0.66	63%	0%
4	Bilateral (above-knee), left side (Ruhe 2004)	0.62	0.47	67%	
	Bilateral (above-knee), right side (Ruhe 2004)		0.55	69%	+2%
5	Unilateral (above-knee), sound side (James and Oberg 1973)	0.94	0.62	65%	
	Unilateral (above-knee), prosthetic side (James and Oberg 1973)		0.68	57%	- 8%
6	Unilateral (above-knee), sound side (Macfarlane et al. 1997)	1.03	0.66	70%	
	Unilateral (above-knee), prosthetic side, (Macfarlane et al. 1997)		0.69	63%	- 7%
7	Unilateral below-knee), sound side (Macfarlane et al. 1991)	1.07	0.67*	67%	
	Unilateral (below-knee), prosthetic side (Macfarlane et al. 1991)		0.65*	63%	- 4%
8	Unilateral (below-knee), sound side (Isakov et al. 1997)	1.33	0.71	69%	
	Unilateral (below-knee), prosthetic side (Isakov et al. 1997)	1.36	0.75	67%	- 2%
9	Unilateral (below-knee), sound side (Bateni and Olney 2002)	1.13	0.67*	64%	
	Unilateral (below-knee), prosthetic side (Bateni and Olney 2002)	1.11	0.66*	60%	- 4%
10	Unilateral (below-knee), sound side (Underwood et al. 2004) SAFE	1.44	0.79	62%	
	Unilateral (below-knee), prosthetic side (Underwood et al. 2004) SAFE		0.82	59%	-3%
11	Unilateral (below-knee), sound side (Underwood et al. 2004) Flexfoot	1.43	0.77	61%	
	Unilateral (below-knee), prosthetic side (Underwood et al. 2004) Flexfoot		0.86	60%	-1%
12	Partial Foot, sound side (Tang et al. 2004)	0.80 (BF); 0.84 (S); 0.84 (P)	0.47 (BF); 0.49 (S); 0.52 (P)	Gait symmetry between prosthetic and sound side: 0.88 (BF), 0.96 (S), 0.98 (P)	
	Partial Foot, prosthetic side (Tang et al. 2004)	0.80 (BF); 0.86 (S); 0.83 (P)	0.55(BF); 0.58 (S); 0.57 (P)		

Table 1.1: Gait speed and symmetry for varying levels of amputation. Several studies are listed to compare differences in results. Results from Macfarlane, et al. (1991) are mean values for subjects wearing a flex-foot and conventional (SACH or uniaxial) foot while walking on a treadmill. Results from Isakov (1997) are for using a SACH foot, while results for Bateni (2002) are for using a SAFE foot, both relatively flexible feet. For the partial foot measurements (Tang et al. 2004), data to calculate stance and swing phases for partial foot amputations were not given, but gait symmetry data is included. Asterisks (*) mark data in which sound limb step length was shorter than prosthetic limb step length, but these were noted as statistically insignificant by the authors. Sound side % stance phase was always larger than prosthetic side.

This dissertation will study the means by which we are able to modulate and increase our step length, whether it is by inherent, learned, or assisted methods. It is hypothesized that people increase their step length *inherently* by several different means: increasing hip flexion and extension; increasing ankle-foot roll over arc length; increasing stance foot heel rise (or “tipping” about the end of the roll over arc) to increase the contact time of the stance leg and allow further hip flexion and extension; and increasing pelvic rotation. Not all subjects may utilize all of these methods, so being able to *learn* these or possibly even other methods of increasing step length may also help them to take an even longer step length. It may also be the case that, due to stability issues or inability of a prosthetic component to function like the able-bodied counterpart, the “normal” methods of regulating step length are not observed. These subjects may then take a shorter step length or compensate in other ways to maintain a longer step length.

1.1 Why study step length?

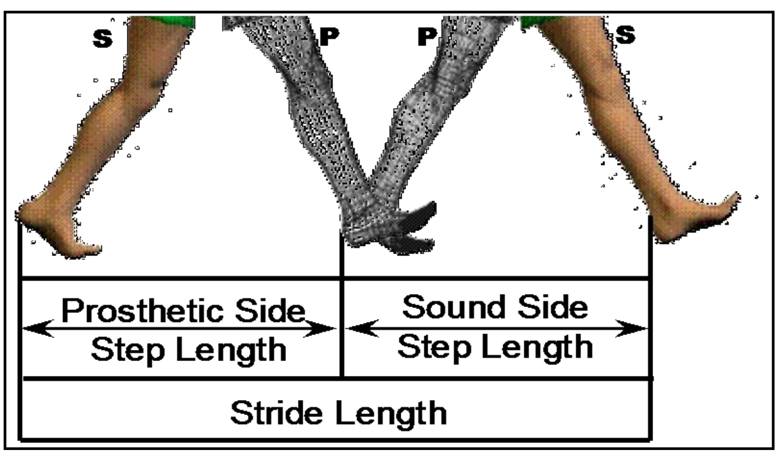

Figure 1.1: Visual depiction of differences in step length between the prosthetic (P) and sound (S) leg of a person with a unilateral lower limb amputation. The sound limb will typically have a shorter step length than the prosthetic side, and usually a shorter swing phase.

Persons with lower limb amputations tend to have slower walking speeds than able-bodied persons, and persons with unilateral lower limb amputation show asymmetry between their sound and prosthetic side step lengths. A list of speeds and step length differences for able-bodied persons and for persons with a range of lower limb amputation levels are compiled in Table 1.1. The stance time on the prosthetic side is usually less than that of the sound limb, and thus swing time on the sound side is shorter than that of the affected side. Along with the differences in percent stance phase, we observe that the step length of the sound side is usually shorter than that of the prosthetic limb (

Figure 1.1). There are many reasons why increased step length would be beneficial for gait. For persons with unilateral lower limb amputations, changes in step length can help make the step cycles between the left and right sides more symmetrical. This may help to decrease risk factors associated with gait asymmetry (Giakas et al. 1996; Horvath et al. 2001; Skinner and Effeney 1985), increase the aesthetics of the person's gait, and increase the distance traveled for a given number of steps.

In a study done by Underwood et al. (2004) the average difference between the sound and prosthetic side step length for eleven individuals having a unilateral below-knee amputation and wearing a SAFE prosthetic foot was 3.0 cm. If symmetry in step length were restored, the increase in distance traversed over 2000 steps (the approximate number of steps in a mile) would be around 30 meters, or just over a quarter length of a football field. Similarly, for bilateral amputees or able-bodied persons, if step length were increased by a seemingly negligible 1.0 centimeter (approximately 1.3% increase in step length, assuming the average normal step length is 0.75 meters) on both sides, one would be able to walk 20 meters further over a period of 2000 steps. This would be about 1/20th the length of an outdoor track (typically 400 meters) (Figure 1.2). The changes would be even more dramatic if a person having a bilateral lower limb amputation could assume the same step length as that of an able-bodied ambulator. Our lab has measured the typical step length for people with bilateral below-knee amputations to be approximately 0.66 meters, a difference of 9.0 cm from that measured from able-bodied ambulators. If we were able to find a way to "restore" this difference in step-length, a person with a bilateral below-knee amputation would be able to walk 180 meters further over a period of 2000 steps -- over 1.5 football fields in length. Persons with partial foot amputations tend to have an even shorter step length (approximately 0.5 meters, as measured by Tang et al. (2004)). An assistive device that would increase the step length of these ambulators seems necessary. In able-bodied gait, step-length increases could be used for obstacle avoidance, or for

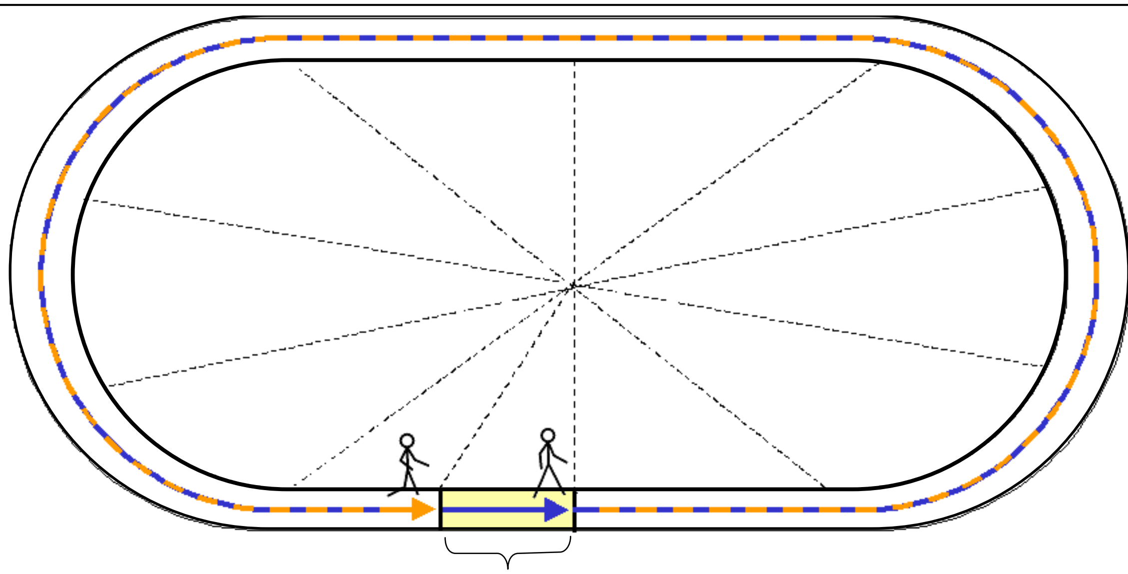

Figure 1.2: By increasing one's step length by a seemingly small 1.0 cm, one can walk 20 meters, or 1/20th the length of a track further over the course of 2000 steps (approximately one mile).

increasing speed without increasing cadence. By being able to increase step length, one could travel further for a given number of steps. This would not only be beneficial for normal walking -- if there are minimal disadvantages -- but it would also help speed or race walkers, or those participating in similar athletics. A method to analyze step length would allow step length analysis and improve upon gait training.

1.2 Hypotheses

This study aims to examine the contributing factors of the lower limb to step length and the methods by which we modulate our step length during walking. The differences between lower limb contributions to step length of different subject groups will also be analyzed. It is believed that all the lower limb segments contribute to the length of a step. For able-bodied persons during freely-selected walking, it is hypothesized that the order these segments contribute to step length (from most to least) are:

- the thigh segments of the trailing and leading limbs,
- the shank segments of the trailing and leading limbs,
- the ankle-foot segment of the trailing limb, and
- the pelvic segment.

For persons who have trained to walk with longer step lengths, the percentage that each segment contributes may be different. For example, persons who race walk may exhibit more pelvic rotation, and thus have a higher contribution to step length by the pelvic segment. Likewise, persons who walk with an amputation may exhibit different segment percentage contributions because their abilities and walking goals may be different than those of able-bodied persons. For example, a shorter prosthetic effective foot length may limit the amount the ankle-foot stance leg segment contributes; or the trailing thigh and shank segment contributions may be limited because the person may feel less stable when extending the trailing limb. Thus, it is hypothesized that persons with lower limb amputation will display different segmental contributions to achieving step length than able-bodied persons, most likely due to a lack in the ability to use their prosthesis in the same way that the intact foot and ankle functions.

1.3 Outline of this dissertation

In order to test these hypotheses, the Segment Contribution to Step Length (SCSL) analysis will be utilized. The SCSL analysis will be introduced in Chapter 2, along with previous research involving step length, and SCSL normative data will be reported (able-bodied subjects walking at their freely-selected walking speed). Chapter 3 will study how the contributions of the lower limb segments change for changing step length of able-bodied persons and determine if gait training has an effect on the lower limb contributions to step length. A comparison will be made between persons trained with race walking techniques and normal able-bodied walking in Chapter 4. This will again explore the effects of gait training on increasing step length. Factors that limit step length in persons with amputation will be explored in the last chapters. Chapter 5 will study the effects that changing the ankle-foot roll over shape arc length has on gait of persons with bilateral trans-tibial amputation. The SCSL analysis will be performed on this group and compared with the able-bodied group in Chapter 6, and differences between these groups will be discussed. In Chapter 7, two case studies will be performed on persons with partial foot amputation walking barefoot and with an orthosis to determine how assistive devices affect gait. A SCSL analysis will be performed in these subjects with partial foot amputation in Chapter 8. Finally, Chapter 9 will discuss the results of the SCSL analysis for all these subject groups, its limitations, and its implications on gait. The culmination of these studies will also help us determine what the main methods are of increasing step length and possible factors that make it difficult to take larger steps than we usually observe. Overall, a better idea of why we walk the way we do and how we might improve gait will be established.

Chapter 2: Step Length of able-bodied self-selected walking and the Segment Contribution to Step Length (SCSL) Analysis

"Bipedalism is a tremendous adaptation for humans and a distinguishing characteristic between humans and other primates" (PBS 2001). The actual reason why we walk bipedally is unknown, but it allows us to carry things such as food, weapons, or tools while allowing us to see farther and walk efficiently. For animals that walk bipedally, a *step* is defined as the sequence of movements the lower limbs take from the time one limb contacts the ground to the time the opposite limb contacts the ground (Figure 2.1). A left and right step together constitutes one *stride* and represents a gait cycle. *Step length* is defined as the distance from a given point on the ipsilateral (originating) foot at initial contact, to a corresponding point of the contralateral (opposite) foot at initial contact.

Step length can vary considerably from person to person. Height, walking speed, and type of gait that a person uses are just some factors that have an influence on step length (Murray et al. 1964; Murray et al. 1970; Perry 1992). Typically, step length is about 0.75 meters for healthy adults walking at their freely selected walking speed of about 1.4 meters/second (Drillis 1958; Murray et al. 1964; Murray et al. 1970; Perry 1992). Many healthy ambulators are capable of producing a step length greater than this in a walking gait, depending on their height and leg length, among other factors. In an observation of 936 people walking down a New York street, Drillis (1958) measured a range of step lengths between 0.54 and 0.99 meters for freely-selected walking.

2.1 Previous research

Human gait has been studied throughout history, with written documents dating as far back as the Egyptians and involving persons such as Leonardo da Vinci and Galileo Galilei. Though step length is often measured and reported during gait analysis studies, only a few investigations have been specifically designed to study step length. Danion et al*.* (2003) studied stride length and frequency variability. They concluded that fluctuations in both step length and cadence increased when either of these parameters were different from that of freely-selected walking, though this variation decreased for increasing stride length. Varraine et al*.* (2000) focused on how stride length adaptation

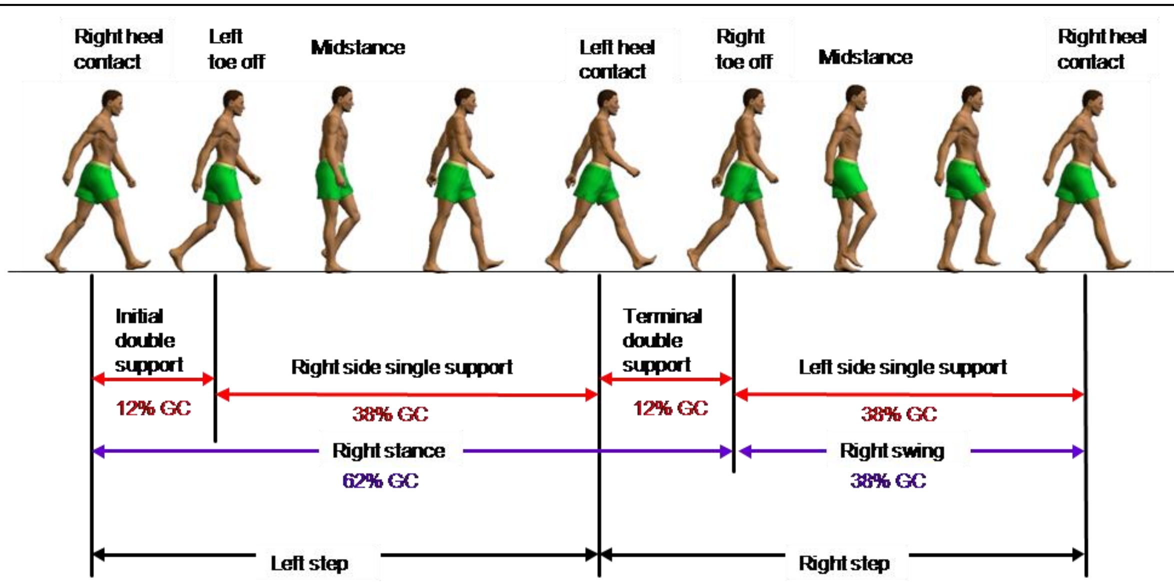

Figure 2.1: Diagram of a typical gait cycle from right heel contact to right heel contact for freely-selected walking. A left and right step make up a stride.

to environmental constraints was controlled. They concluded that active deceleration of the swing leg by the hip extensors allowed stride shortening while stride lengthening was controlled by an increase in activity of the ankle and hip of both legs. An analysis of joint angles also found that as step length increased, the maximum extension angle increased at the hip and ankle of the trailing limb. Grieve (1968) introduced the speed (v) - step length (SL) power law relationship as: $SL \sim v^{0.42}$, which was assumed to be the step length humans tend to choose to minimize metabolic energy consumption for a given speed. Others have derived different power law ratios that are dependent on the square root of speed (Miff 2000; Milner and Quanbury 1970).

Several studies have analyzed the influence of step length and cadence on gait characteristics (Kirtley et al. 1985; Laurent and Pailhous 1986; Murray et al. 1966; Nilsson and Thorstensson 1987). A few have specifically examined the differing effects between increasing step length and increasing cadence (Miff 2000; Nilsson and Thorstensson 1987). Other studies involving step length have investigated its relationship with ground reaction forces (Martin and Marsh 1992; Soames and Richardson 1985); swing and stance phases (Milner and Quanbury 1970); vertical displacement of the trunk (Miff et al. 2001); and metabolic cost (Bertram and Ruina 2001; Kuo 2001; Kuo et al. 2005; Molen et al. 1972). These studies suggest that freely-selected walking speeds, step lengths, or cadences are chosen to optimize certain factors such as reduce energy cost or lower vertical displacement of the trunk. Lower limb ranges of motion to changing step length were studied by Ohmichi and

Miyashita (1983) and Zarrugh and Radcliffe (1979). Ohmichi and Miyashita acquired data from two subjects, analyzing data from each subject separately. Pelvic rotation in the transverse plane was reported for only a single subject. They reported that step length changes from 0.5 to 1.0 meters saw increases from 5° to 19° of pelvic rotation. Zarrugh and Radcliffe found a similar increase in pelvic rotation for a subject walking at free and fixed walking trials of around 5° to 25° for step lengths between 0.64 and 1.0 meters.

With advances in technology of motion acquisition and analysis, the study of step length and gait can be more comprehensively analyzed than in previous studies. Specifically, the contributions of the lower limb segments to step length can be better analyzed. Better generalized conclusions of the effects of step length on walking will improve our understanding of gait and can help us determine where differences occur in step length between able-bodied persons and persons with gait pathology.

2.2 The Segment Contribution to Step Length (SCSL) Analysis

Step length is an important parameter to report in gait analyses because it can be used as a comparative tool in determining functionality. Joint kinematics, normally in the form of joint angles, are typically reported as part of a quantitative gait analysis, and these are also used to compare different walking conditions or subject groups. Trying to relate the contributions of the joints of the two legs and the pelvis to step length using angle measurements can be confusing because their contributions are dependent upon both the segment lengths and orientations in space.

The Segment Contribution to Step Length (SCSL) analysis is intended to simplify the assessment of the various factors affecting step length by directly calculating the contribution of each lower limb segment to the overall step length. Reported measurements using the SCSL analysis would combine segment length and orientation into one measurement. This would allow us to more easily determine how the contributions of the lower limb segments to step length are changing when different walking conditions or subject groups are compared. One can determine what kinematic changes of the lower limb contribute to the overall step length. It can also be determined if an overall change in segment contribution is changing a person's step length, or if particular segments modulate the overall step length. Using the SCSL analysis will also help to determine if different subject groups modulate their step length in different ways.

The SCSL analysis looks at the contribution of six lower limb segments to the overall step length (Figure 2.2). These segments are: 1) the trailing ankle-foot complex, 2) the trailing shank, 3) the trailing thigh, 4) the pelvis, 5) the leading thigh, and 6) the leading shank. To calculate the contribution of each segment, the fore-aft distance moved by each of these segments was calculated. For the ankle-foot segment, segment contribution was measured as the fore-aft distance the ankle moves from ipsilateral initial foot contact to contralateral initial foot contact. For the other five segments (trailing shank, trailing thigh, pelvis, leading shank, and leading thigh,) segment contribution was measured as the fore-aft distance of these segments at time of contralateral initial foot contact.

Segments were measured as the distance between joint centers which were calculated during a static standing trial using OrthoTrak software (Motion Analysis Corporation, Santa Rosa, CA). The ankle-foot segment was measured as the distance moved for

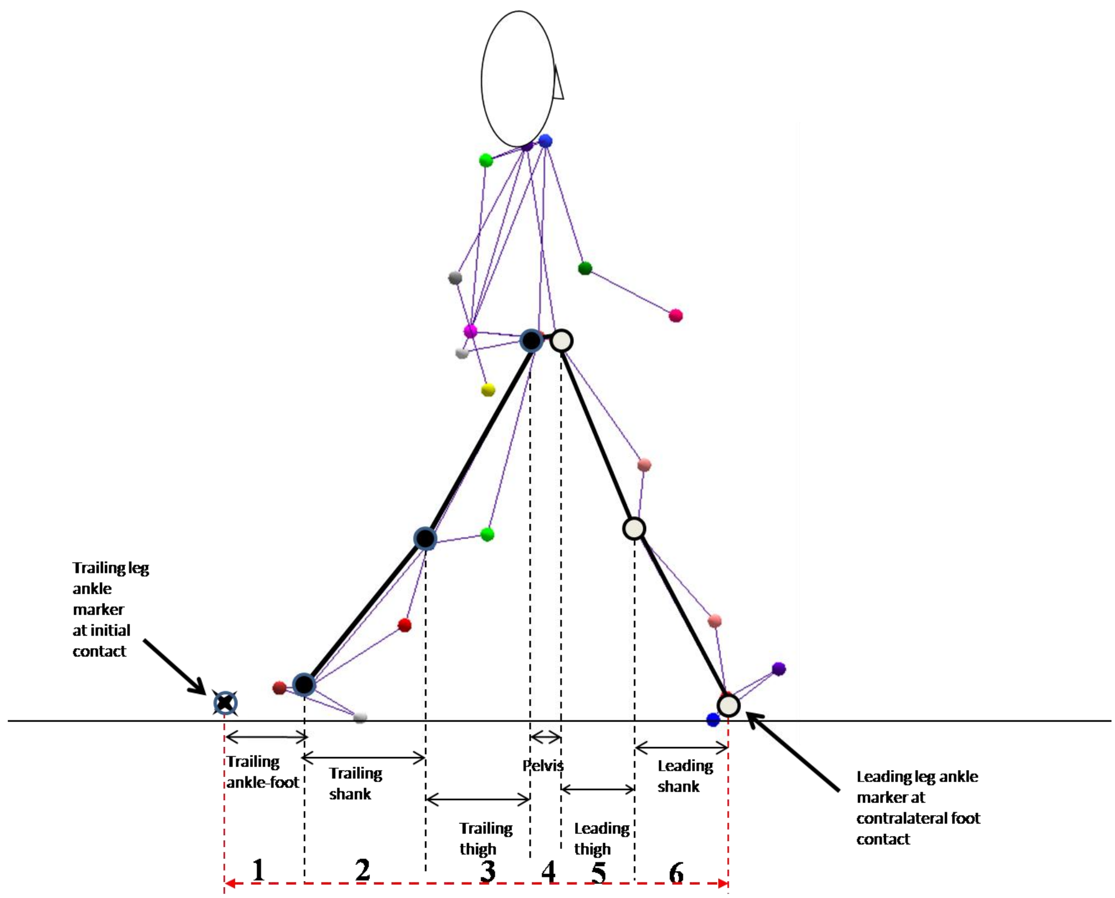

Figure 2.2: Stick figure made up of the 23 markers placed on each subject during walking trials (using a modified Helen Hayes marker set). Bold black lines outline the six lower limb segments used for the SCSL analysis (numbered in bold). End points for each segment are located at joint center positions of the ankles, knees, and hips. The trailing ankle-foot segment is the sagittal distance the ankle joint center moves from ipsilateral heel contact to contralateral heel contact. The other segments (segments 2-6) are measured at time of contralateral heel contact.

the trailing ankle joint center from ipsilateral foot contact to contralateral foot contact. The shank segment was defined as the segment between the ankle joint center and the knee joint center. The thigh segment was defined as the segment between the knee joint center and the hip joint center, and the pelvic segment was defined as the segment between the left and right hip joint centers.

Overall step length was measured as the fore-aft distance between the trailing leg's ankle joint center at time of initial foot contact to the leading leg's ankle joint center at contralateral foot contact. Step length was calculated from the ankle joint centers to eliminate differences in foot marker placement of the different study populations.

These measurements were normalized to eliminate effects of leg length of each subject by dividing each measurement by his or her leg length. Leg length was defined as the vertical distance of the ASIS markers to the ground when the subject was standing upright (taken during a static trial). The percentage contribution of each segment to overall step length was also calculated, which was derived by dividing the measured data by the overall step length. A study was performed with able-bodied individuals to determine the contribution of each lower limb segment to step length to establish normative data and to create a control measurement for comparison with other populations. It was hypothesized that the order these segments contribute to step length (from most to least) are:

- the thigh segments of the trailing and leading limbs,
- the shank segments of the trailing and leading limbs,
- the ankle-foot segment of the trailing limb, and
- the pelvic segment.

2.3 Methods

Preliminary analysis suggested that ten subjects were needed to determine differences in joint angles and segment contributions to step length equivalent to one standard deviation of the measurements, assuming allowable type I error (α) of 5% and type II error (β) of 20% (statistical power of 80%) (Lieber 1990). Subjects signed consent forms that were approved by Northwestern University's Institutional Review Board. Data collection and analyses for the study were conducted in the VA Chicago Motion Analysis Research Laboratory (VACMARL). Wearing athletic shoes, subjects were asked to walk across a flat walkway at their freely-selected normal walking speed while kinematic and kinetic data were acquired using an eight-camera Eagle Digital Real-Time motion measurement system (Motion Analysis Corporation, Santa Rosa, CA) at 120 Hz and six AMTI (Advanced Mechanical Technology, Inc., Watertown, MA) force platforms at 960 Hz. A modified Helen Hayes marker set (Kadaba et al. 1990) was used to define a biomechanic model on each person. A static standing trial was performed before the walking experiment in order to estimate the location of the joint centers of rotation. Walking trials were repeated until 3-5 clean force platform hits were obtained for each leg. A clean hit is defined as one in which one foot contacts the force plate and stays within the bounds of the plate, and the other foot does not also contact the same plate during the walking trial.

2.4 Data Analyses

Data were processed using Motion Analysis' EVa and Orthotrak software. Missing data points in marker position data were interpolated using a cubic spline technique. Raw marker position data were filtered using a fourth-order bidirectional Butterworth infinite-impulse response digital filter with an effective cutoff frequency of 6.0Hz. Data were further processed using custom macros in Microsoft Excel (Microsoft Corporation, Redmond, WA) and Matlab (The Mathworks, Inc., Natick, MA). Specific gait data that were analyzed for this study included temporospatial data and joint angles. A Segment Contribution to Step Length (SCSL) Analysis was also performed. Data were normalized by leg length to eliminate effects of longer leg lengths by some subjects. The percentage contribution to overall step length was also calculated for each segment. Sagittal plane kinematics were obtained for comparison with previous research.

2.5 Results

2.5.1 Vital Statistics and Temporospatial Parameters

Table 2.1: Subject's vital statistics and temporospatial information

Subject	Age	Gender	Weight (kg)	Height (cm)	Leg length (cm)	Mean step length (cm)	Mean speed (cm/sec)
1	26	M	86.0	191.5	109.0	76.9	128.7
2	23	F	60.0	171.0	94.0	67.9	130.2
3	24	F	57.5	165.0	91.0	77.7	161.0
4	24	M	70.5	170.0	103.0	72.3	122.3
5	24	F	57.8	171.0	98.0	69.0	137.2
6	23	F	65.0	164.0	95.0	69.5	130.7
7	24	M	71.5	180.5	104.0	72.6	129.5
8	27	F	81.0	168.5	94.5	70.3	132.0
9	26	M	70.0	178.0	102.0	76.1	141.5
10	26	M	73.4	186.0	108.0	79.6	131.6
Average	**24.7**	**-**	**69.3**	**174.6**	**99.9**	**73.3**	**134.5**
St Dev	**1.4**	**-**	**9.5**	**9.1**	**6.2**	**4.3**	**10.6**

The group of subjects consisted of 5 males and 5 females. Vital statistics as well as step length and mean speed are reported in Table 2.1. Leg length was between 55-60% of overall height. Mean normalized step length was 0.74 times leg length (Figure 2.3) with the exception of Subject 3 whose step length was around 85% of her leg length. Average freely-selected walking speed for the group of 10 subjects was measured to be 1.35 m/s, mean cadence was about 114 steps/minute, and mean double support time as a percentage of the gait cycle was 11% while mean stance time was 61% of the gait cycle. These values of temporospatial data are within

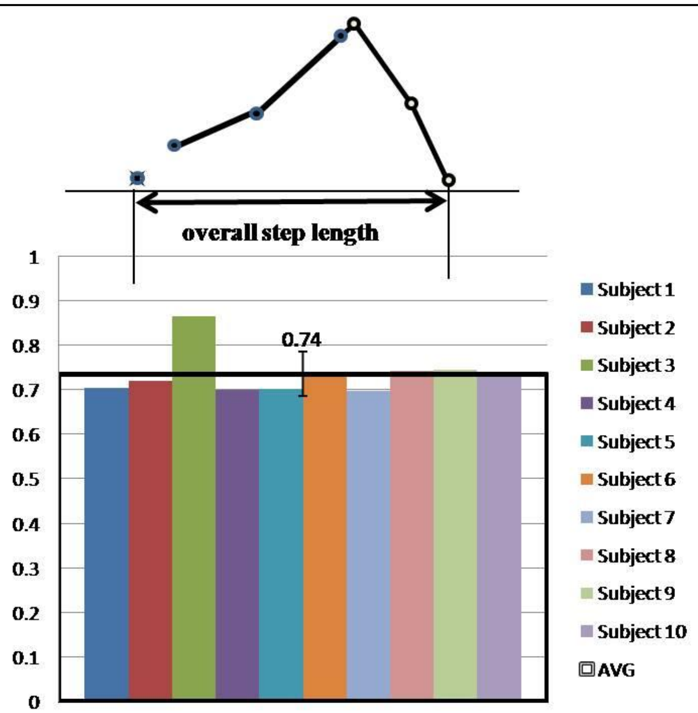

Figure 2.3: Step length of all ten subjects normalized by leg length (LL). Mean value of all ten subjects is represented by the black horizontal line across the graph. Standard deviation is plotted as the black vertical line on the average line between Subjects 5 and 6. Mean step length was measured as 0.74 x LL.

range of that reported from previous studies of able-bodied persons walking at freely-selected walking speeds.

2.5.2 **Kinematics**

Mean sagittal joint kinematics of the ankle, knee, and hip, along with pelvic rotation over the gait cycle was measured for the 10 subjects (Figure 2.4). Since this group walked fairly symmetrically, values were measured and reported for the left side only. Maximum mean ankle dorsiflexion of 12° occurred at 50% of the gait cycle while maximum ankle plantarflexion of 14° occurred at 68% of the gait cycle (during swing). Maximum mean knee flexion of 58° occurred at 75% of the gait cycle, while minimum knee flexion of -8° occurred at 98% of the gait cycle. Maximum mean stance phase knee flexion for the 10 subjects was measured to be 13° at 18% of the gait cycle. Maximum hip flexion of 28° occurred at 54% of the gait cycle while minimum hip flexion of -14° occurred at 87% of the gait cycle. Minimum pelvic rotation of -2° occurred at 23% of the gait cycle while maximum mean pelvic rotation of 3° occurred at 72% of the gait cycle. Total mean sagittal plane ankle, knee, and hip range of motion (ROM) over the gait cycle was 26°, 65°, and 41°, respectively, while mean pelvic rotation ROM was 6°.

Angle measurements were also made at the beginning of the gait cycle, at the time of ipsilateral initial contact for the left side. Mean values for ankle, knee, and hip flexion

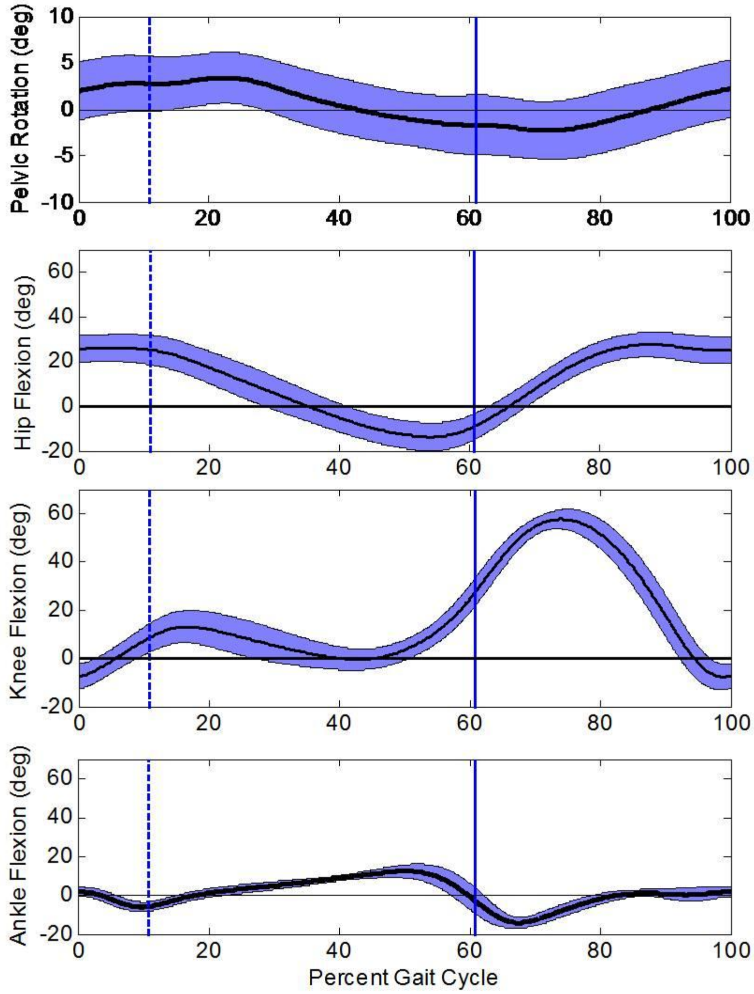

Figure 2.4: Mean ankle, knee, and hip sagittal plane kinematic and pelvic rotation angles for ten able-bodied subjects walking at freely-selected walking speed over a gait cycle. Shaded area indicates one standard deviation. Dotted vertical line represents time of contralateral toe-off. Solid vertical line represents ipsilateral toe-off.

angles were 2°, -7°, and 26°, respectively. Transverse pelvic rotation angle at the beginning of the gait cycle was measured to be 2°.

2.5.3 Segment Contribution to Step Length Analysis

Segment contributions to step length of the six lower limb segments are graphed in Figure 2.5 through Figure 2.10. Reported data were normalized by leg length. Data was reported in this way to determine the similarities in contribution of each segment by the able-bodied subjects. Bars represent each subject, and the solid horizontal line represents the average value of all 10 subjects. Mean contribution by the trailing ankle-foot segment to overall step length was 0.11 times leg length (LL). Contribution by the stance shank and thigh were measured to be 0.16 LL and 0.15 LL, respectively. Pelvic contribution to step length was positive for some subjects and negative for others, with mean contribution to pelvic segment averaging out to 0.00 LL. Actual values were small, ranging between -0.008 and 0.012 LL. Contribution by the leading thigh and shank segment to overall step length was 0.14 LL and 0.18 LL, respectively. From the graphs, it was observed that, when normalized by leg length, variability between subjects was fairly low, particularly for the ankle-foot and shank segments.

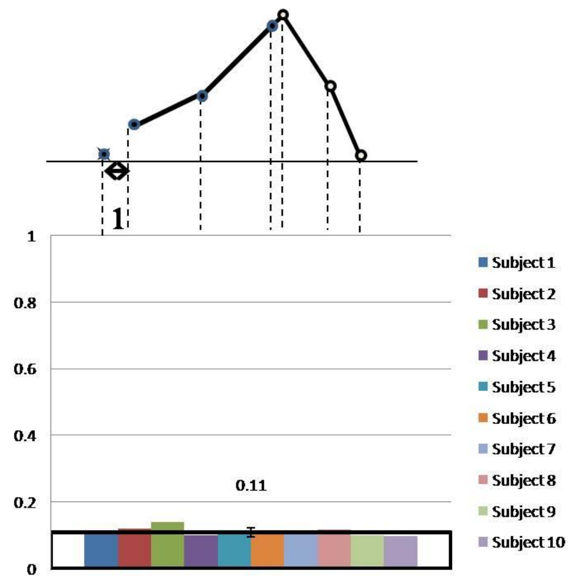

Figure 2.5: Trailing leg ankle-foot segment contribution (normalized by leg length) to overall step length. Mean value of all ten subjects is represented by the black horizontal line across the graph. Standard deviation is plotted on the average line between Subject 5 and Subject 6. Mean contribution by the trailing ankle-foot segment to overall step length was 0.11 times leg length.

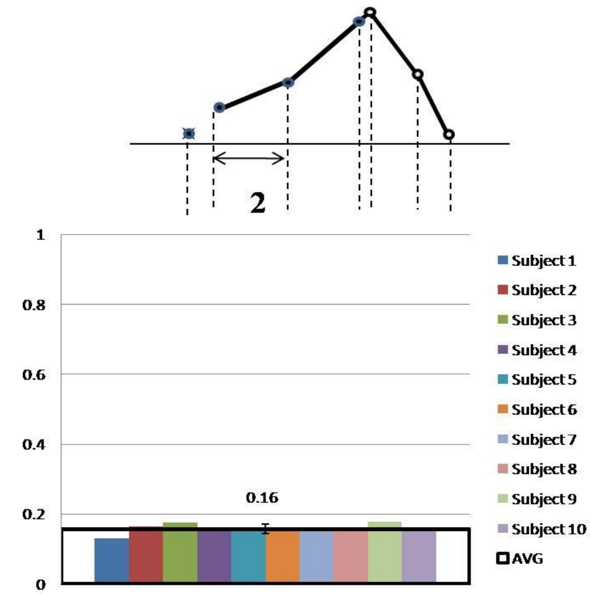

Figure 2.6: Trailing leg shank segment contribution (normalized by leg length) to overall step length. Mean value of all ten subjects is represented by the black horizontal line across the graph. Standard deviation is plotted on the average line between Subject 5 and Subject 6. Mean contribution by the trailing shank segment to overall step length was 0.16 times leg length.

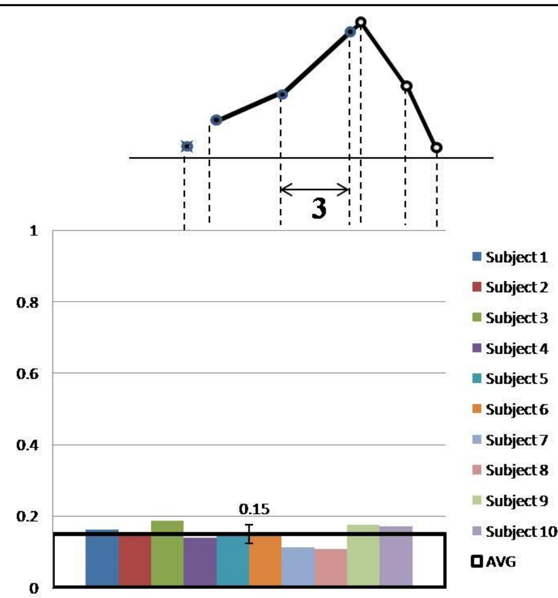

Figure 2.7: Trailing leg thigh segment contribution (normalized by leg length) to overall step length. Mean value of all ten subjects is represented by the black horizontal line across the graph. Standard deviation is plotted on the average line between Subject 5 and Subject 6. Mean contribution by the thigh segment to overall step length was 0.15 times leg length.

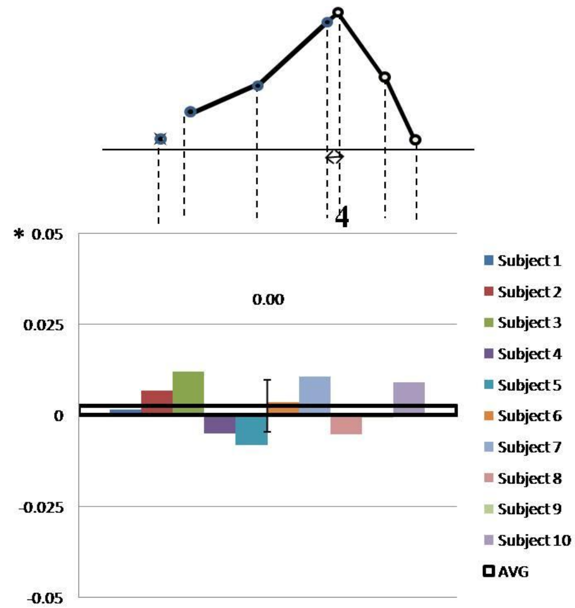

Figure 2.8: Pelvis segment contribution (normalized by leg length) to overall step length. Mean value of all ten subjects is represented by the black horizontal line across the graph. Standard deviation is plotted on the average line between Subject 5 and Subject 6. Mean contribution by the pelvic segment to overall step length was 0.00 times leg length. *Note that the axes are different than the other segments because of the very small contribution values.

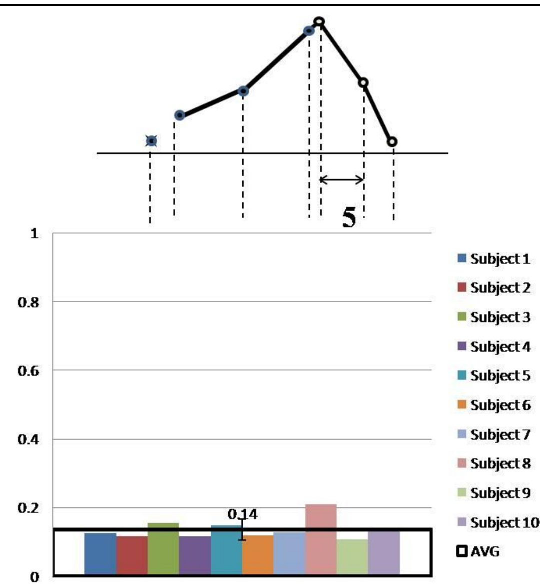

Figure 2.9: Leading thigh segment contribution (normalized by leg length) to overall step length. Mean value of all ten subjects is represented by the black horizontal line across the graph. Standard deviation is plotted on the average line between Subject 5 and Subject 6. Mean contribution by the thigh segment to overall step length was 0.14 times leg length.

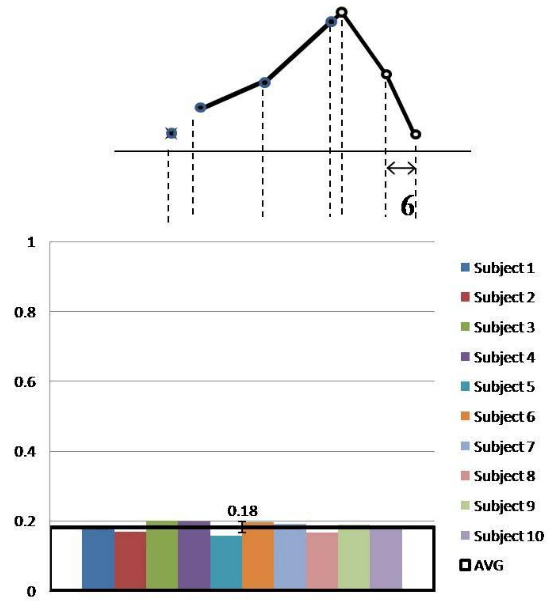

Figure 2.10: Leading shank segment contribution (normalized by leg length) to overall step length. Mean value of all ten subjects is represented by the black horizontal line across the graph. Standard deviation is plotted on the average line between Subject 5 and Subject 6. Mean contribution by the thigh segment to overall step length was 0.18 times leg length.

Contribution by each segment as a percentage of overall step length was also measured (Figure 2.11). Leading and trailing shank segments contributed the largest amount to overall step length (24.8% and 21.4%, respectively,) followed by the trailing and leading thigh (20.3% and 18.4%, respectively,) the trailing ankle-foot (14.8%) and a small percentage by the pelvis (0.3%). The leading leg (made up of the leading thigh and shank segments) contributed 43.2% while the trailing limb (made up of the trailing ankle-foot, shank, and thigh segments) contributed 56.5% to overall step length.

2.6 Discussion

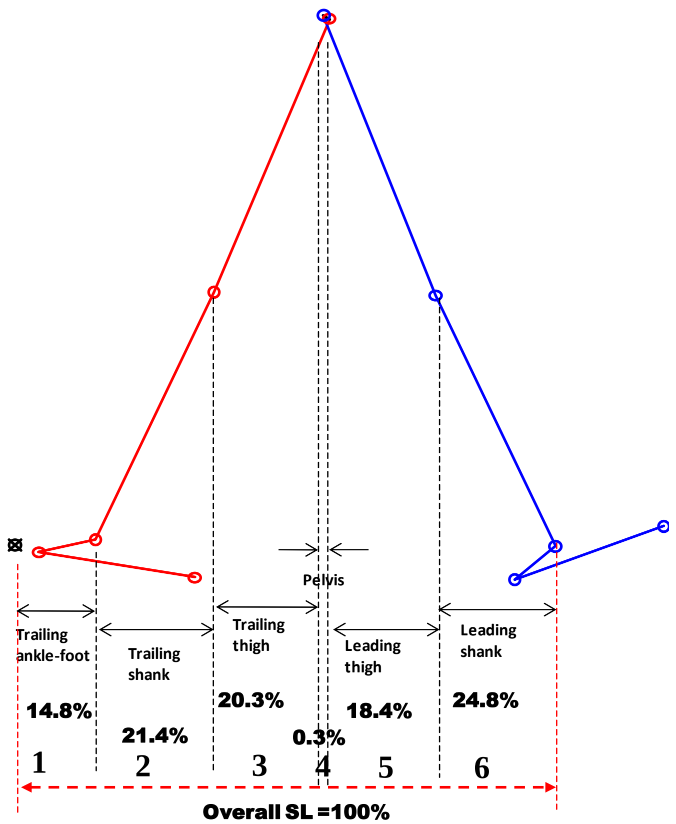

Figure 2.11: Percent contribution of each of the six lower limb segment to overall step length. Accompanying stick figure is of a typical able-bodied subject's lower limb joint centers (and heel and toe markers) at the end of a step.

The sagittal plane ankle and hip flexion, and pelvic rotation kinematics of these subjects are similar to those reported in previous studies of able-bodied persons (Murray 1967; Murray et al. 1964; Perry 1992). Sagittal plane knee flexion kinematics displayed more knee hyperextension at initial foot contact (-7.6°) than that normally observed, but is still within range of that noted by Perry (1992).

Leg length normalized segment contributions to step length were fairly similar between subjects. Except for the pelvic segment (for which relative contribution was fairly small), the coefficient of variation (standard deviation divided by mean, times 100) was between 8% and 22%. For freely-selected walking, 80% of able-bodied subjects' step length is contributed by the leading and trailing shank and thigh segments, almost split equally (20%) by each of the four segments. The trailing limb contributed about 13% more than the leading limb for freely-selected walking, with the contribution by the trailing ankle-foot being the cause of this increase. Surprisingly, pelvic rotation does not contribute much to step length for normal walking (0.3%).

This normative data would be a useful tool for determining differences between subject groups or types of walking. One could use this tool to distinguish differences in lower limb segment contributions for changes in step length which affects walking speed. For example, SCSL comparisons with this control group and a group of persons with lower limb amputation may help us determine if the decrease in step length is due to fit or function of the prosthetic device. It might also be observed that step length differences are due to changes in contributions by the residual limb segments due for other reasons such as for increased stability or because of changes in physical ability and joint ranges of motion.

Chapter 3: Step length modulation and increasing step length of able-bodied persons

Although step length is often measured during gait analysis studies, there has been little research specifically studying the modulation of step length. It is known that increased walking speeds of most people naturally occur by increasing both step length and cadence in a nearly linear fashion (Inman et al. 1981; Koopman 1989; Milner and Quanbury 1970; Waters et al. 1988). Power law ratios dependent on the square root of speed have been reported by Miff (2000) and Milner and Quanbury (1970). The method by which step length increases has not previously been described in detail. For step length changes due to adaptation on different terrains, stride lengthening is controlled by an increase in “propulsive” forces (increased activity of the ankle and hip extensor muscle) of the stance leg and an increase in swing duration of the contralateral leg (Varraine et al. 2000). Joint angle analysis determined that, for increasing step lengths, the maximum extension angle increases at the hip and ankle of the stance leg. A study by Milner and Quanbury (1970) measured the timing of different phases of the gait cycle for varying speeds and concluded that as walking speed increases (from 0.61 m/s to 2.04 m/s) the double support percentage of gait decreases (from 30% to 18%) while swing percentage increases (from 35% to 41%). Other studies have looked at lower limb range of motion (ROM) changes for changing step length (Ohmichi and Miyashita 1983; Zarrugh and Radcliffe 1979) during free and fixed speed walking. Results from these studies suggest that increases in pelvic rotation and hip and ankle flexion ROM are likely methods used to increase step length (Figure 3.1 and Figure 3.2). Though analysis of joint angles can help determine how changes to step length are occurring, it is difficult to determine to what extent each lower limb segment contributes to these changes because an association must be made between both segment lengths and their orientations in space.

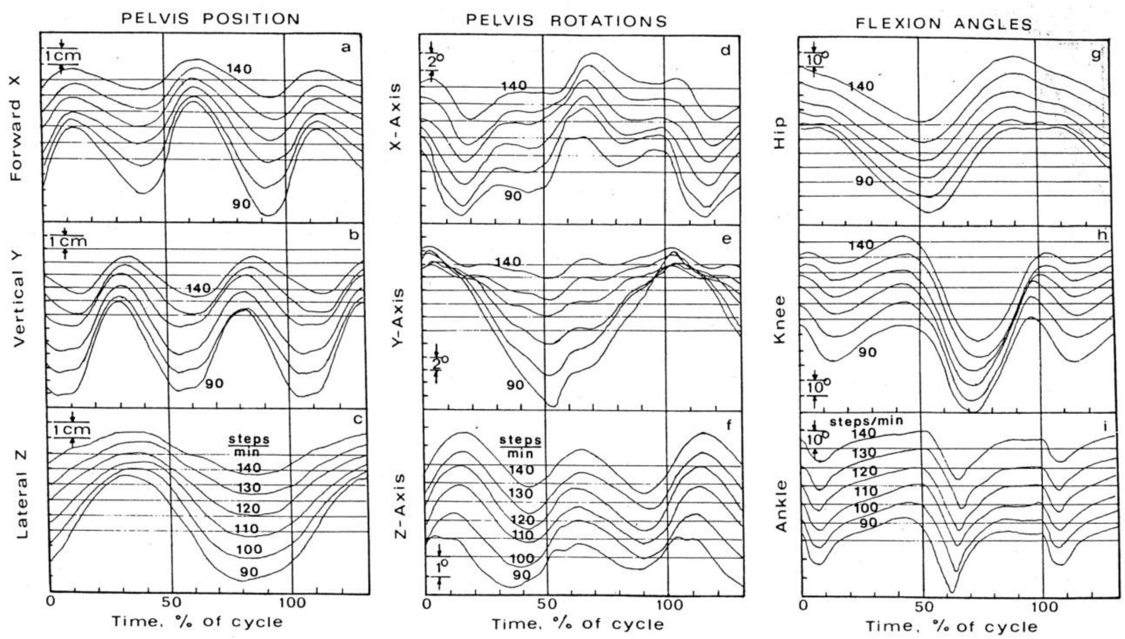

Figure 3.1: Position (left) and rotation (middle) of the pelvis, and flexion angles for hip, knee, and ankle (right) for fixed velocity walking of 1.5 m/s with changing cadence (and thus step length) as measured by Zarrugh and Radcliffe (1979). Note the differences in scale between the pelvic rotation graphs (middle) and the flexion angles of the other joints (right).

With advances in technology of motion acquisition and analysis, the study of increasing speed via changes in step length can be more comprehensively examined than in previous research. This study aims to analyze the methods for modulating step length of able-bodied persons. It will also be determined if step length can be increased by training techniques. A Segment Contribution to Step Length (SCSL) analysis will be performed to better understand how each of the lower limb segments contributes to step length increases. Kinematics and ankle-foot roll over shapes will be analyzed. By looking at the changes in segment contributions to step length as the step length changes in able-bodied persons, we can determine if there is a general pattern followed by all able-bodied persons. Knowing the individual components that make up the total step length may help to provide insight into gait therapies for persons with disabilities or may lead to new prosthetic or orthotic designs that can improve upon step length and gait. Better generalized conclusions of the effects of step length on walking will be obtained to come to an improved understanding of gait.

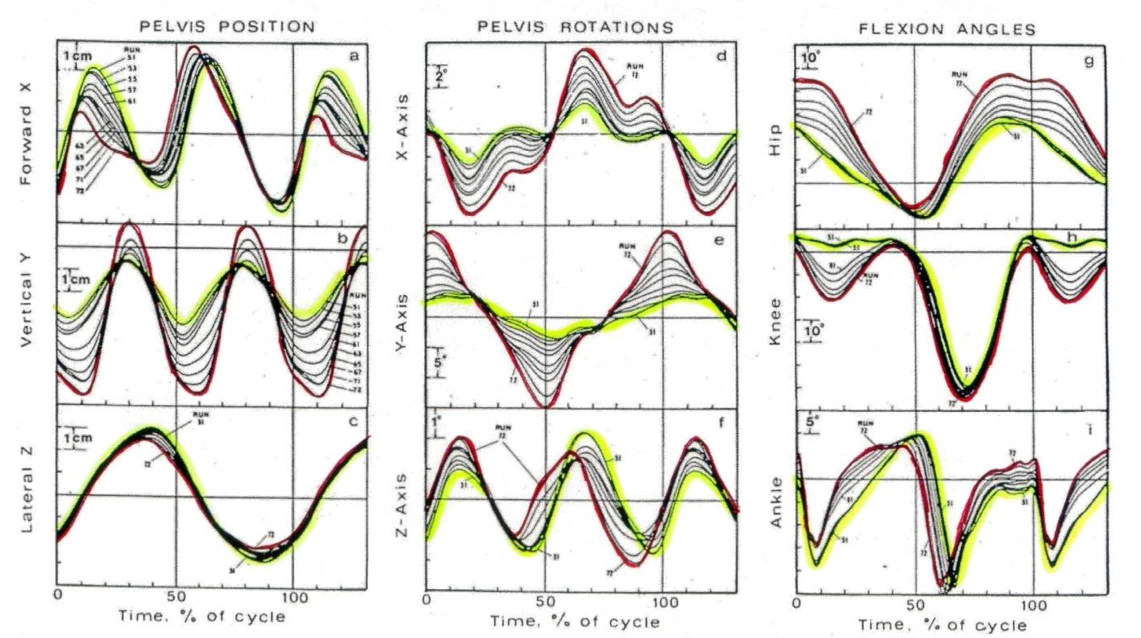

Figure 3.2: Position (left), rotation (middle) of the pelvis, and flexion angles of hip, knee, and ankle (right) for increasing speed as measured by Zarrugh and Radcliffe (1979). The fastest (red) and slowest (yellow) speeds are highlighted for clarification. Pelvic translation and rotation, and hip, knee, and ankle flexion increased for increasing speeds. Note the differences in scale between the pelvic rotation graphs (middle) and the flexion angles of the other joints (right).

3.1 Methods

3.1.1 Gait data acquisition

Preliminary statistical analysis indicated that ten subjects were needed to determine differences in segment contributions equivalent to one standard deviation of the measurements, assuming allowable type I error (α) of 5% and type II error (β) of 20% (statistical power of 80%) (Lieber 1990). Subjects signed consent forms that were approved by Northwestern University's Institutional Review Board. Subjects wore athletic shoes and comfortable clothing during the study. Data collection and analyses for the study were conducted in the VA Chicago Motion Analysis Research Laboratory (VACMARL). An eight-camera Eagle Digital Real-Time motion measurement system (Motion Analysis Corporation, Santa Rosa, CA) was used to acquire marker movements at 120 Hz and calculate kinematic data. Ground reaction forces were acquired using six AMTI (Advanced Mechanical Technology, Inc., Watertown, MA) force platforms simultaneously recorded with the motion analysis cameras at 960 Hz. A modified Helen Hayes marker set (Kadaba et al. 1990) was used to define a biomechanic model of each person. A static standing trial was performed before the walking experiment in order to estimate the location of the joint centers of rotation.

A total of nine different walking trials were performed by each subject. The first six conditions were to study gait at different step lengths. The last three conditions were to study the effects of gait training on step length. Subjects were first asked to walk at their freely-selected walking speed across the walkway (labeled as walking condition 2 since the actual measured step length fell between 0.65 m and 0.87 m). They then walked at four other set step lengths (cadence was self-selected) which were indicated by markers and lines projected on the ground. These step lengths in order were:

- 0.65 m (walking condition 1),
- 0.87 m (walking condition 3),
- 1.09 m (walking condition 4), and
- 1.4 times leg length (1.4xLL) (walking condition 5) which was chosen because pilot studies suggested this was the longest step length able-bodied persons are able to take.

Leg length was measured as the distance from the right ASIS to the ground.

After the first five walking conditions were performed, subjects were asked to take the longest step length possible (walking condition 6) across the walkway. No marks or lines were placed on the ground for this or subsequent walking trials. Walking conditions 7-9 were then performed to determine if training suggestions enabled each subject to take even longer step lengths. These suggestions were those used in race walking techniques. Subjects were allowed to practice until they were comfortable, which was approximately 3-5 minutes. These three walking conditions were:

- longest step length possible while emphasizing rotation of the pelvis in the transverse plane (walking condition 7),

- longest step length possible with emphasis on keeping the trailing foot on the ground as long as possible, and allowing smooth rollover from heel to toe, (walking condition 8), and
- longest step length possible after practicing these two training techniques (walking condition 9).

Walking trials were repeated until 3-5 clean force platform hits were obtained for each leg. Subjects were allowed to rest in between trials as needed.

3.1.2 Data analyses

Data were processed using Motion Analysis' EVa and Orthotrak software. Missing data points in marker position data were interpolated using a cubic spline technique. Raw marker position data were filtered using a fourth-order bidirectional Butterworth infinite-impulse response digital filter with an effective cutoff frequency of 6.0Hz. Data were further processed using custom programs in Microsoft Excel (Microsoft Corporation, Redmond, WA) and Matlab (The Mathworks, Inc, Natick, MA). It was assumed that subjects walked fairly symmetrically on both sides. Data for step length analysis were calculated from left side steps.

Specific gait data that were analyzed for this study were walking speed, step length, cadence, and single and double limb support times for all walking conditions, as well as sagittal plane ankle, knee, and hip flexion, and transverse pelvic rotation kinematics. A Segment Contribution to Step Length (SCSL) analysis was also performed. The SCSL analysis calculates the contribution of six lower limb segments to the overall step length.

The segments that were included were the trailing ankle-foot, shank, and thigh segments, the pelvic segment, and the leading thigh and shank segments. (The details about how each segment contribution is calculated are provided in Chapter 2.) Data were normalized by leg length (LL) to eliminate effects of longer leg lengths of some subjects. The percentage contribution to overall step length was also calculated for each segment. Ankle-foot roll over shapes and effective foot length ratios (EFLR) were also calculated. Roll over shapes are created by transforming center of pressure data from the laboratory-based coordinate system to a body-based coordinate system (Hansen, Childress et al. 2004) (see Appendix). The EFLR is a measure of the distance the COP progresses under the foot and is calculated as the length of the roll over foot shape (i.e. the distance from the heel to the anterior end of the shape) divided by the total foot length (Hansen, Sam et al. 2004).

3.1.3 **Statistical methodology**

Mean data during each walking condition for each subject was obtained. The relationship between step length and the percent contribution of each of the six lower limb segments was investigated using the Pearson product-moment correlation coefficient. This correlation technique tells us how well a linear relationship fits the two variables being measured. The Pearson correlation coefficient (r) has a value between -1 to +1, indicating whether there is a negative or positive correlation, respectively. The size of the absolute value indicates the strength of the relationship, with 1 being a perfect linear correlation and 0 indicating no linear correlation between the two variables. The strength of correlation was determined to be small (weak) if the absolute value of r was between 0.10 to 0.29; medium if it was between 0.30 to 0.49 and large (strong) if r was between 0.50 to 1.0 (Cohen 1988).

The Pearson correlation does not take into account that some of the data points are dependent, since there are ten subjects walking at nine different step length conditions. Therefore, a one-way repeated measures ANOVA was also performed to compare how segment contribution to step length varied between walking conditions 1 through 5 (step lengths between 0.65m and 1.4 times leg length). Mauchly's Test of Sphericity was performed to test assumptions of the ANOVA test. When the assumption of sphericity was violated, Greenhouse-Geisser correction factor was used to determine the P value. Pairwise comparisons were performed using Bonferroni adjustments when the data was found to be significant. Values for P were adjusted by the software to reflect the Bonferroni correction (i.e. adjusting the cut-off P values to 0.05).

SPSS software (SPSS Inc, Chicago, IL) was used to perform the statistical analyses, and the level of statistical significance for each test was set at a value of $P<0.05$. Data were found to be normally distributed using the Shapiro-Wilk Test of Normality.

3.2 Results

For a more informative analysis of the data, walking conditions for some results are reported in two groups:

1. Increasing step length walking conditions (walking conditions 1-6), and

2. Longest step length conditions, pre and post training (walking conditions 5-9)

In this way we can determine 1) what changes occur for increasing step length and 2) what affect training might have on gait when subjects were asked to walk at their longest step length.

3.2.1 Subject information and temporospatial data

Table 3.1: Subjects' vital statistics

Subject	Gender	Age	Weight (kg)	Height (cm)	Leg length (cm)
1	M	26	86.0	191.5	109.0
2	F	23	60.0	171.0	94.0
3	F	24	57.5	165.0	91.0
4	M	24	70.5	170.0	103.0
5	F	24	57.8	171.0	98.0
6	F	23	65.0	164.0	95.0
7	M	24	71.5	180.5	104.0
8	F	27	81.0	168.5	94.5
9	M	26	70.0	178.0	102.0
10	M	26	73.4	186.0	108.0
Average	**-**	**25**	**69.3**	**174.6**	**99.9**
St Dev	**-**	**1**	**9.5**	**9.1**	**6.2**

The group of subjects consisted of 5 males and 5 females. Vital statistics are reported in Table 3.1. Average age, weight, and height were 25 years, 69.3 kg, and 174.6 cm, respectively. Leg length was between 55-60% of overall height. Table 3.2 lists the mean temporospatial measurements for the ten subjects at

Walking Condition	Walking Description (SL=Step Length)	Speed (m/sec)	Step Length (m)	Cadence (steps/min)	Double support time (% of gait cycle)	Stance time (% of gait cycle)
1	SL=0.65m	1.14	0.66	114	13%	63%
2	Normal Walking	1.35	0.73	110	11%	61%
3	SL=0.87m	1.47	0.86	104	10%	60%
4	SL=1.09m	1.76	1.05	99	8%	58%
5	SL=1.4 times leg length	2.02	1.42	87	5%	54%
6	SL = longest (pre coach)	2.01	1.37	89	6%	56%
7	SL=longest, stance foot on ground	2.13	1.31	95	6%	56%
8	SL=longest, coach pelvic rotation	2.08	1.36	96	7%	57%
9	SL=longest (post coach)	2.22	1.36	99	6%	56%

Table 3.2: Description of each walking condition and mean speed, step length, and cadence for ten able-bodied subjects walking at these walking conditions. For walking conditions 1 through 5, marks were placed on the ground at the measured step lengths. For condition 5, leg length was measured as the vertical distance from the ASIS to the ground. Subjects walked fairly symmetrically, so opposite foot contact occurred at 50% of the gait cycle for all walking conditions.

each of the walking conditions. Mean step length is also plotted in Figure 3.3. Step lengths ranged between 0.66 and 1.42 m. Longest mean step length occurred for walking condition 5 (step length of 1.4 times leg length). Fastest walking speed of 2.22 m/s occurred for walking condition 9, which was 0.87 m/s faster than freely-selected walking. As step length increased, double support and stance times became shorter.

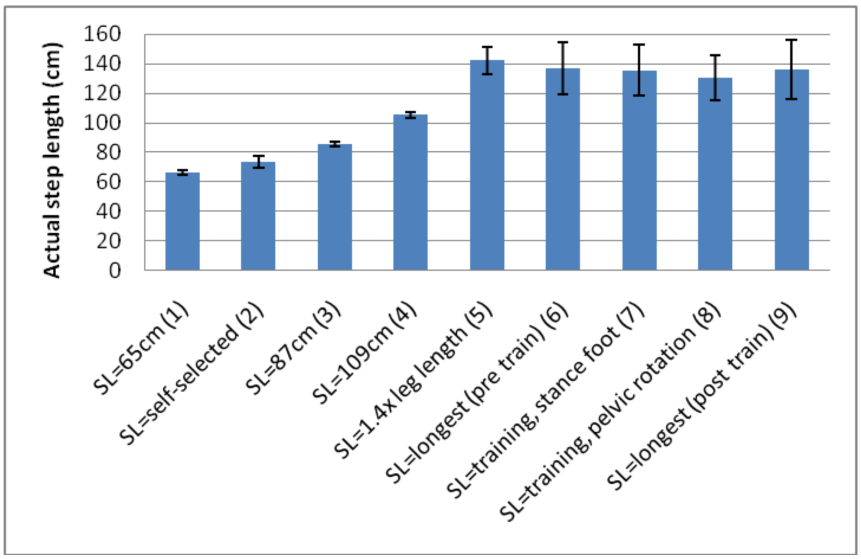

Figure 3.3: Mean step length and standard deviation of 10 subjects for each of the 9 different walking conditions.

3.2.2 Kinematics

Sagittal plane ankle, knee, and hip angles, as well as pelvic rotation in the transverse plane over the gait cycle are plotted in Figure 3.4 for walking conditions 1 though 6. The largest mean pelvic rotation range of motion (ROM) of almost 40° occurred for the longest mean step length under walking condition 5 (SL=1.4 times leg length). This is more than a 30° increase in ROM compared to that for freely-selected walking. An increase in hip flexion peak from 25° to almost 70° during initial stance phase was observed, with walking condition 1 (SL=0.65m) having the lowest peak value as well as the lowest overall ROM. Though hip flexion ROM increased by 57° peak to peak

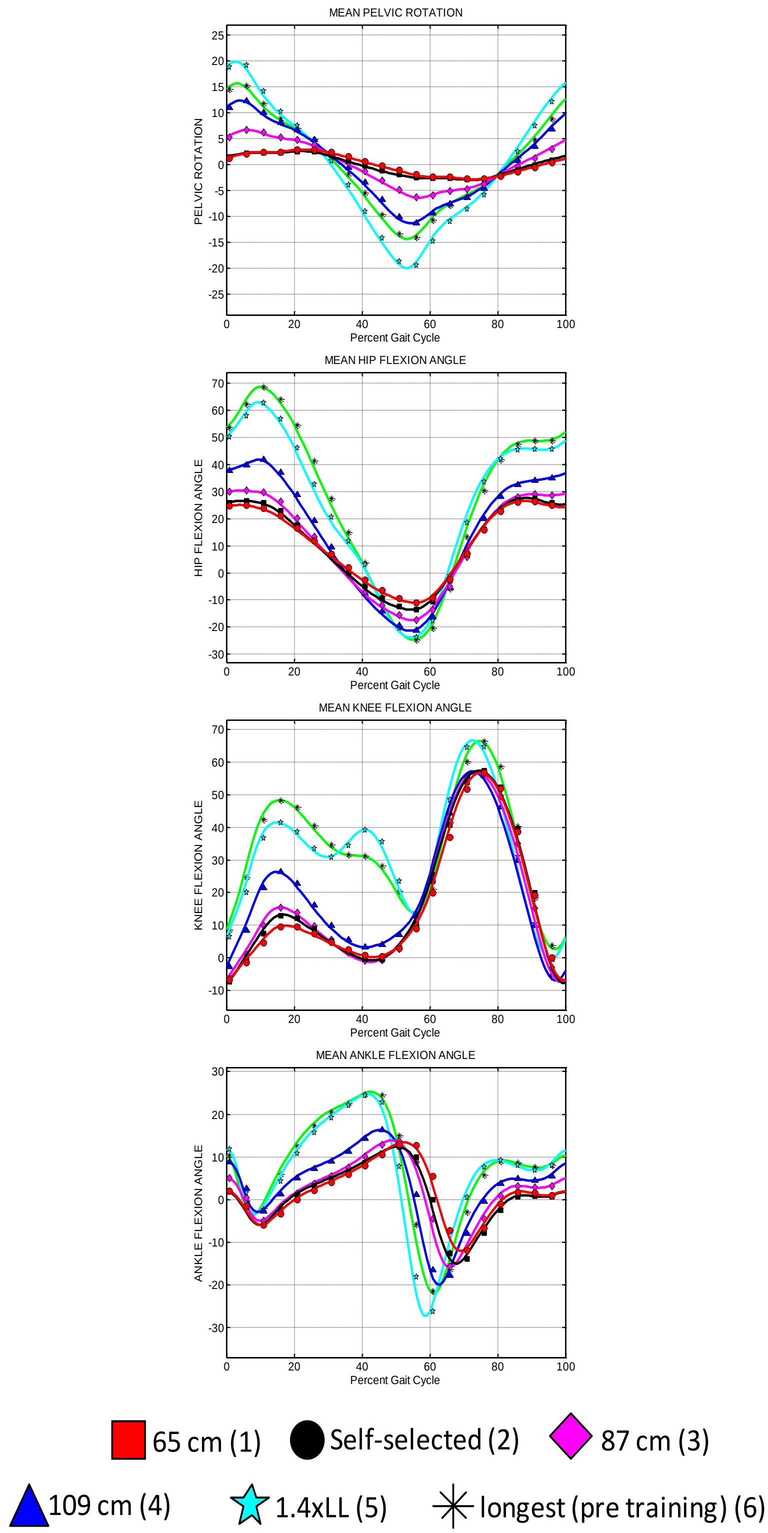

Figure 3.4: Mean sagittal plane ankle, knee, and hip, and pelvic rotation angles over the gait cycle of 10 able-bodied subjects for six different step length walking conditions (conditions 1-6).

between the shortest step length (condition 1) and the longest possible step length before training (condition 6), peak hip extension increased by less than 15°. This suggests that, when taking longer steps, the leading limb extends further forward than the trailing limb which is also observed in the SCSL analysis. For increasing step lengths, hip flexion peaks tend to occur after contralateral foot contact.

Increase in knee flexion over the stance phase of gait was observed to occur with increasing step length. An increase in peak stance phase knee flexion of almost 40° for large step lengths was also observed. For walking conditions 5 and 6 (1.4 times leg length and longest step length before training, respectively) the knee remained flexed by at least 20° throughout stance.

As step length increased, ankle dorsiflexion angle at initial foot contact (0% of the gait cycle) increased by approximately 10°. For longer step lengths (greater than 1.09 m), peak ankle dorsiflexion occurred earlier in the gait cycle (approximately 45%) before contralateral foot contact, and increased by approximately 15° (from 12° to 27°) compared to normal walking. Peak ankle plantarflexion also increased from 12° to 29° as step length increased (between conditions 1 and 6) and occurred earlier in the gait cycle, shifting from 70% to 59%, though this peak still occurred during the swing phase of gait.

Only slight changes were observed in kinematics for walking conditions 5-9 (longest step lengths, pre and post training conditions) (Figure 3.5). The largest sagittal plane ankle and knee ROM was observed for walking condition 5 (step length of 1.4 times leg length), though these differences were fairly negligible. Walking condition 8 (training, pelvic rotation,) had the largest pelvic rotation ROM (42° peak to peak), although similar values were also obtained during walking condition 5.

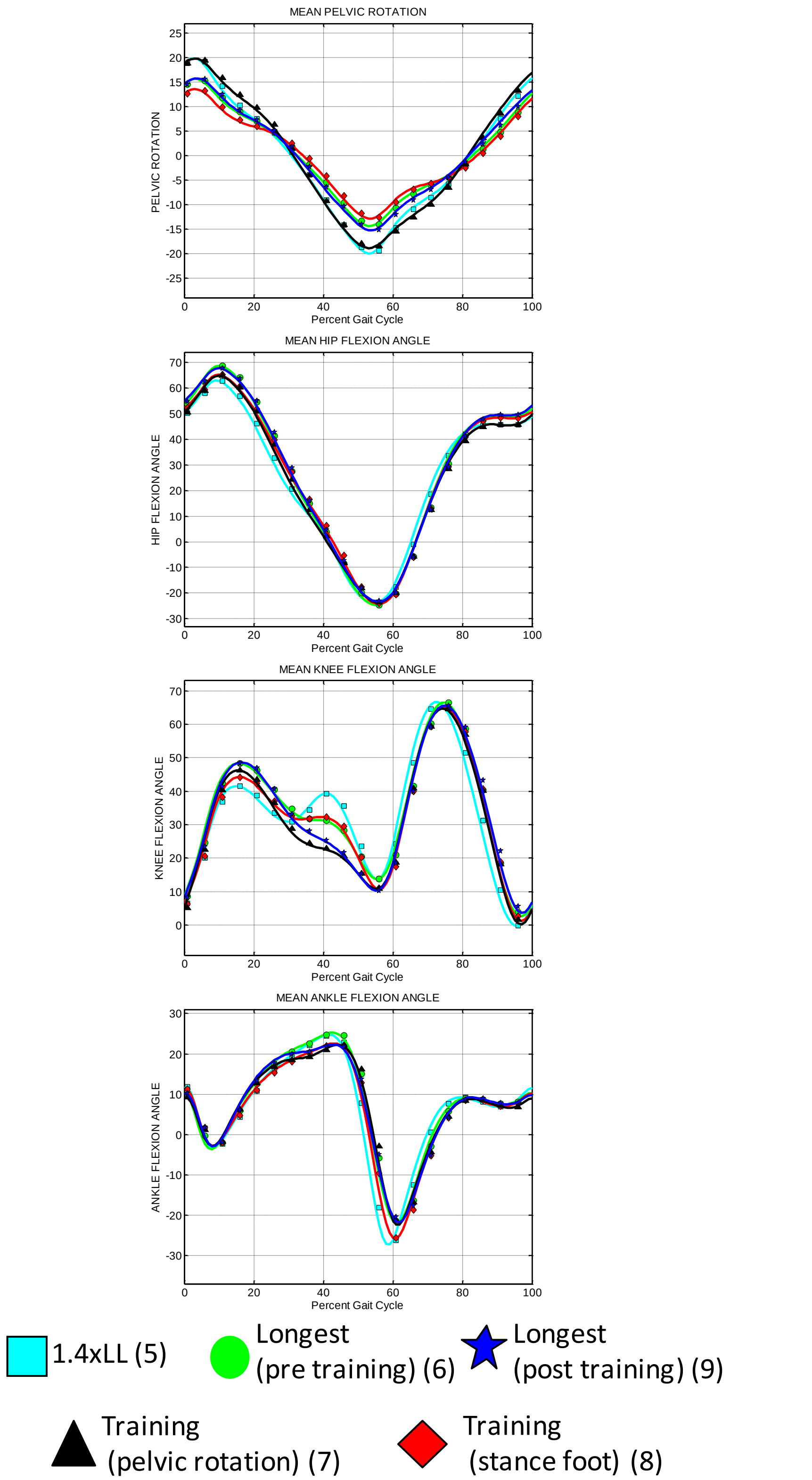

Figure 3.5: Mean sagittal plane ankle, knee, and hip, and transverse pelvic rotation angles over the gait cycle of 10 able-bodied subjects for pre and post training conditions (walking conditions 5-9).

The overall ROM of the lower limb joints over the gait cycle for the nine different walking conditions are plotted in Figure 3.6. As the step length increased for walking conditions 1 through 5, the ROM of all the joints increased, though the change in knee flexion ROM generally due to increases in knee flexion during the swing phase of gait. The knee remained flexed throughout the gait cycle for step lengths greater than

1.09m. A comparison of the joint ROM during the stance phase of gait (Figure 3.7) shows that knee ROM is between 20° and 30° less compared to that obtained over the entire gait cycle. Stance-phase knee ROM generally increased as step length increased, which was also true for the other lower limb joints during the stance phase of gait.

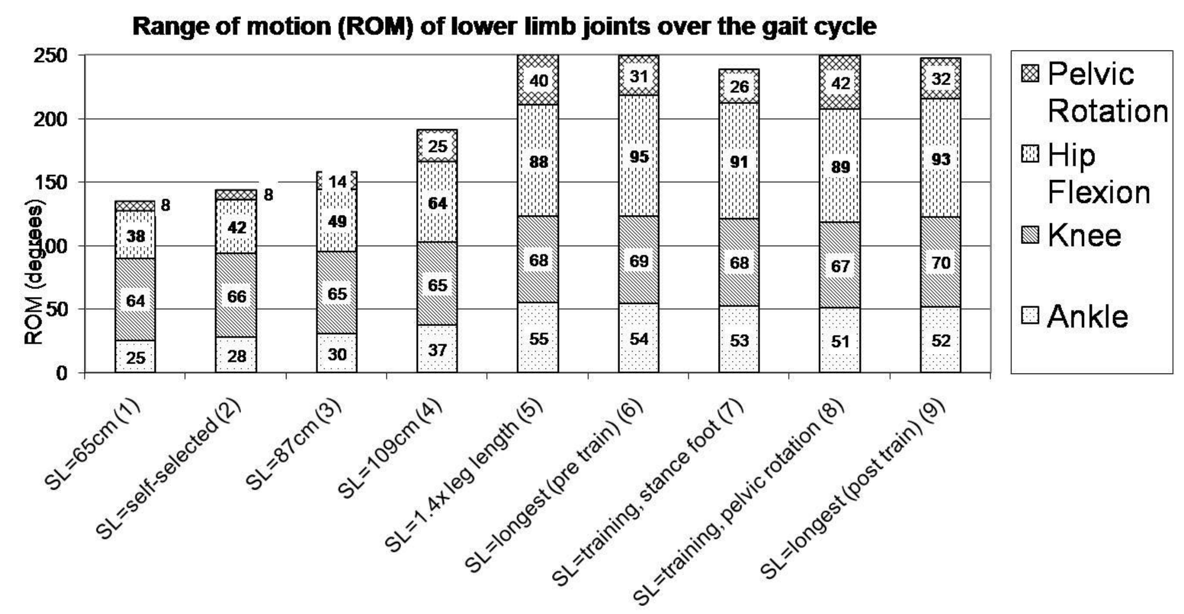

Figure 3.6: Overall mean range of motion (ROM) for hip, knee, and ankle flexion and pelvic rotation over the entire gait cycle for walking conditions 1 through 9.

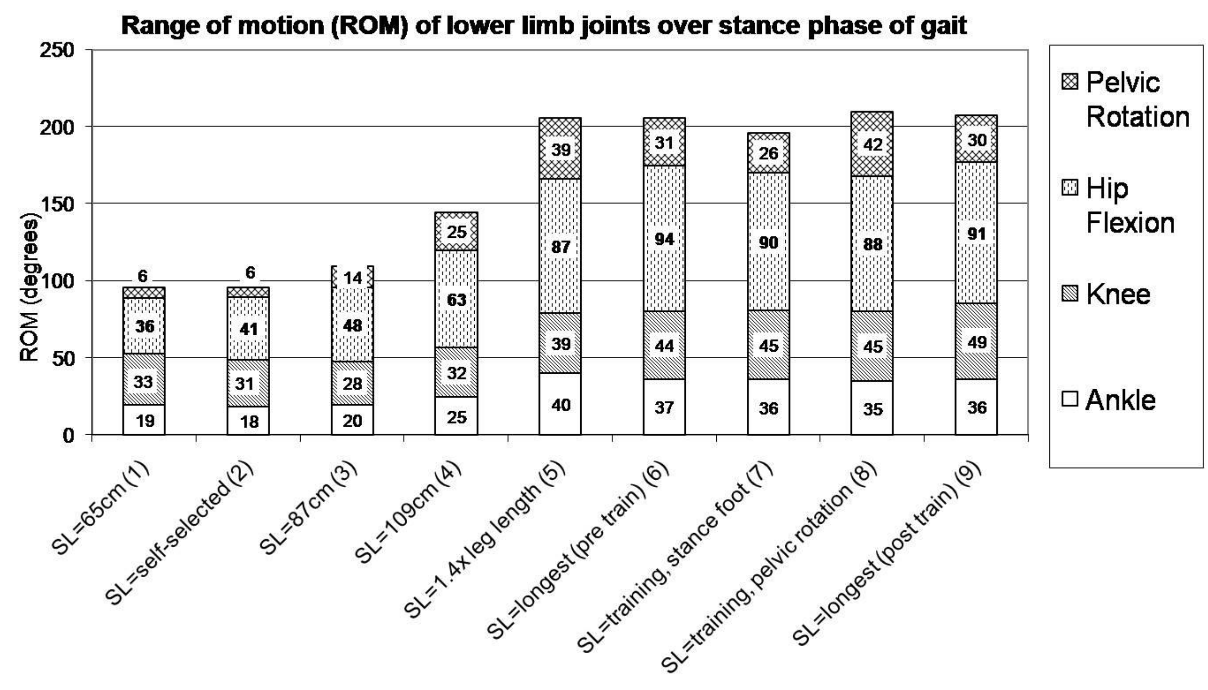

Figure 3.7: Overall mean range of motion (ROM) for hip, knee, and ankle flexion and pelvic rotation over the stance phase of gait for walking conditions 1 through 9.

3.2.3 Segment Contribution to Step Length analysis

Segment contributions to step length of the six lower limb segments are plotted in Figure 3.8 through Figure 3.13. Data are reported both normalized by leg length and as a percentage of the overall step length.

Contribution of the trailing ankle-foot segment displayed a somewhat linear increase from 0.1 to 0.3 cm/leg length (LL) for step length increases of 0.65 to 1.80 m, respectively (Figure 3.8). Though the actual contribution of the ankle-foot segment increased as step length increased, there were only slight changes as a percentage contribution to overall step length, remaining between 10% and 20% of overall step length. This was still measured as a strong, positive correlation in the relationship between step length and percentage contribution of the trailing ankle-foot segment to step length (r=0.64, n=90, P<.0005) using the scale from Cohen (1988). Longer step lengths were associated with higher percentage contributions. The ANOVA results conclude that the percentage contribution to step length by the trailing ankle-foot segment was significantly affected by the step length walking condition, F(1.75,15.73)=19.55, P<0.001 with Greenhouse-Geisser correction. Pairwise comparisons determined that significant differences were only found between condition 5 (step length of 1.4 times leg length) compared with conditions 1 through 4 (step lengths ranging from 0.67 and 1.09 m).

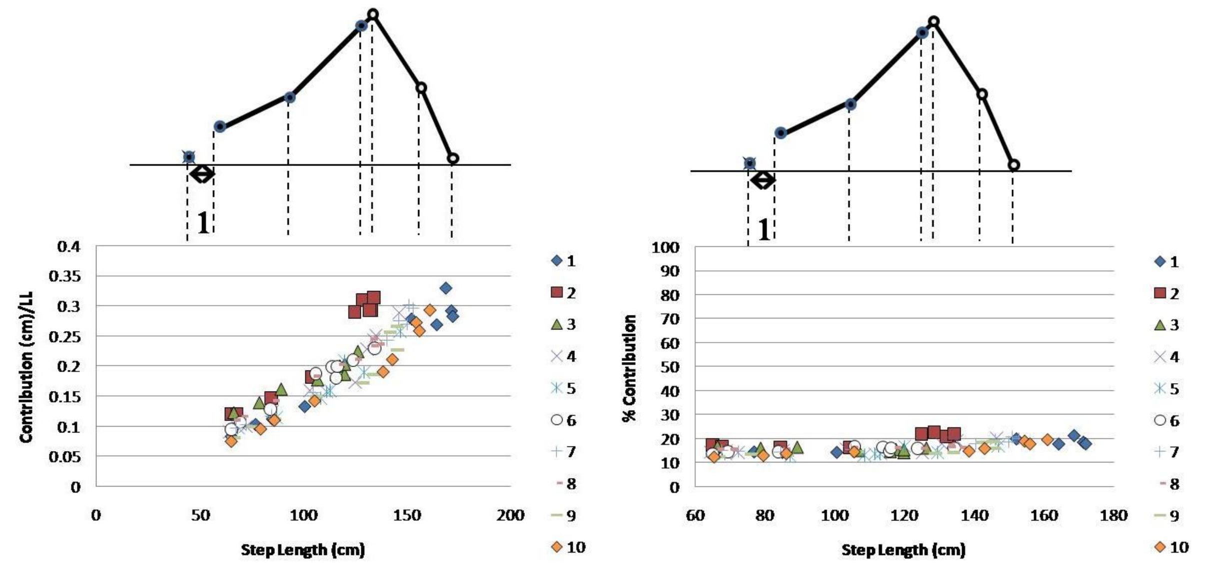

Figure 3.8: Trailing ankle-foot segment contribution to step length versus actual step length for all ten subjects and all walking conditions. Contribution is shown normalized by leg length (left) and as a percentage of overall step length (right).

The trailing shank segment contributed a similar amount to the trailing ankle-foot segment at lower step lengths (0.10 LL), but was able to contribute more at longer step lengths (up to 0.38 LL) (Figure 3.9). As a percentage of overall step length, the trailing shank segment contribution increased slightly (from 20 to 25% of overall step length) as step length increased. This was measured as a strong, positive correlation between step length and percentage contribution of the trailing shank segment to step length (r=0.59, n=90, P<.0005), with longer step lengths associated with higher percentage contribution by the trailing shank segment. ANOVA results concluded that percentage contribution to step length by the trailing shank segment was significantly different between walking conditions 1 through 5, F(2.16,19.47)=9.58, P<0.001 with Greenhouse-Geisser correction. Pairwise comparisons indicate that significant differences were only found between condition 5 (step length of 1.4 times leg length) compared with conditions 1 through 3 (step lengths ranging between 0.67 and 0.87m).

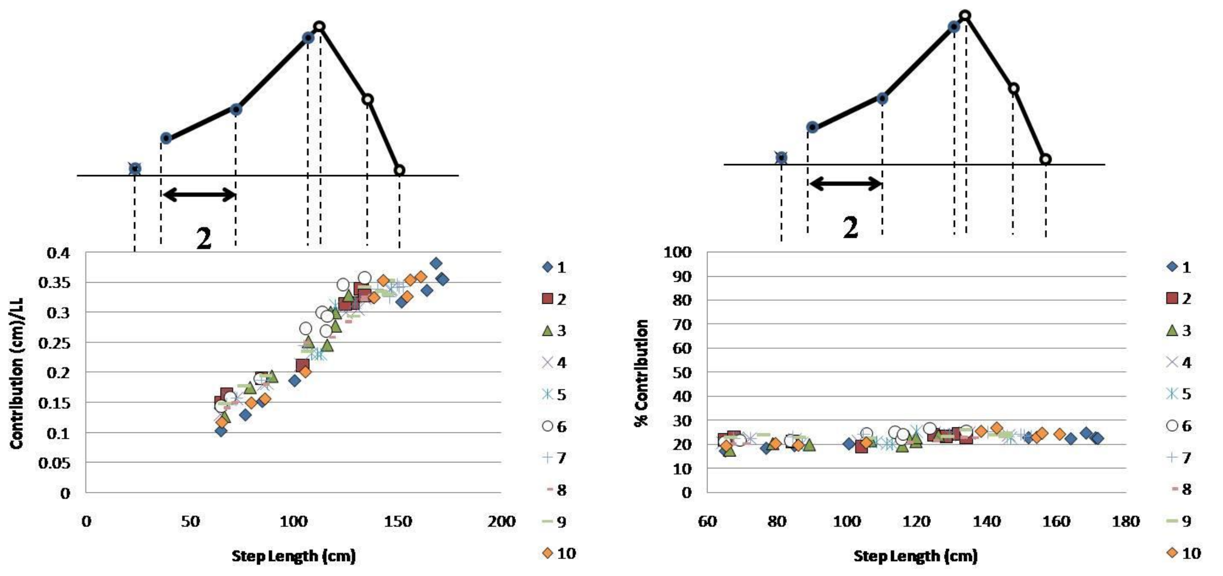

Figure 3.9: Trailing shank segment contribution to step length versus actual step length for all ten subjects and all walking conditions. Contribution is shown normalized by leg length (left) and as a percentage of overall step length (right).

The trailing thigh contribution increased from approximately 0.14 to 0.30 LL for increasing step length, which remained around 20% of overall step length (Figure 3.10). There was a moderate, negative correlation in the relationship between step length and percent contribution of trailing thigh segment to step length (r=-.38, n=90, P<.0005), with longer step lengths associated with a lower percentage contribution by the trailing thigh segment. One-way ANOVA of walking conditions 1 through 5 found that percentage contribution to step length by the trailing shank segment was significantly different between walking conditions, F(2.23,20.08)=8.66, P=0.001 with Greenhouse-Geisser correction. Pairwise comparisons concluded that significant differences exist between walking conditions 1 (step length of 0.67m) and 2 (freely-selected step length); conditions 2 and 5 (step length of 1.4 times leg length); and conditions 3 (step length of 0.87 m) and 5.

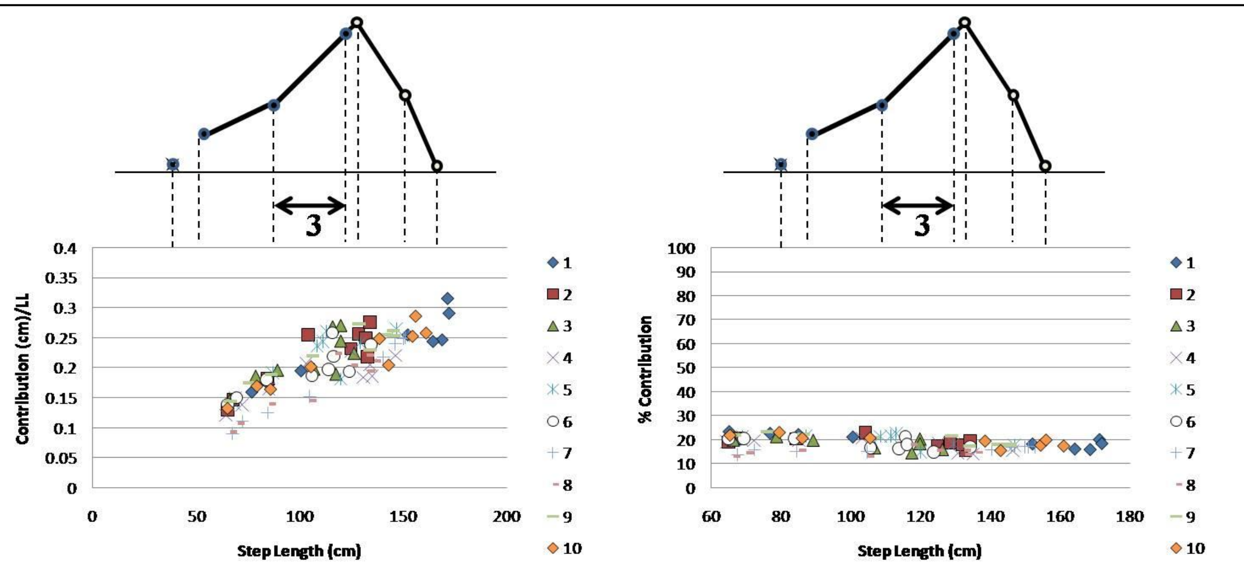

Figure 3.10: Trailing thigh segment contribution to step length versus actual step length for all ten subjects and all walking conditions. Contribution is shown normalized by leg length (left) and as a percentage of overall step length (right).

Though contribution of the pelvic segment to step length increased linearly as step length increased, the contribution was very small compared to other segment contributions (between 0.0 and 0.1 LL) (Figure 3.11). The percent contribution to overall step length was near 0%, and remained under 10% even at the longest step lengths. This was still measured as a strong, positive correlation between step length and percent contribution of the pelvic segment to step length ($r=0.64$, $n=90$, $P<.0005$), according to the scale defined by Cohen (1988). Longer step lengths were associated with higher percentage contribution by the pelvic segment. One-way ANOVA found that percentage contribution to step length by the pelvic segment was significantly different between walking conditions 1 through 5, $F(1.54,13.84)=44.04$, $P<0.001$ with Greenhouse-Geisser correction. Pairwise comparisons determined that significant differences occurred between all walking conditions 1 through 5 except between condition 1 (step length of 0.65 m) and condition 2 (freely-selected step length).

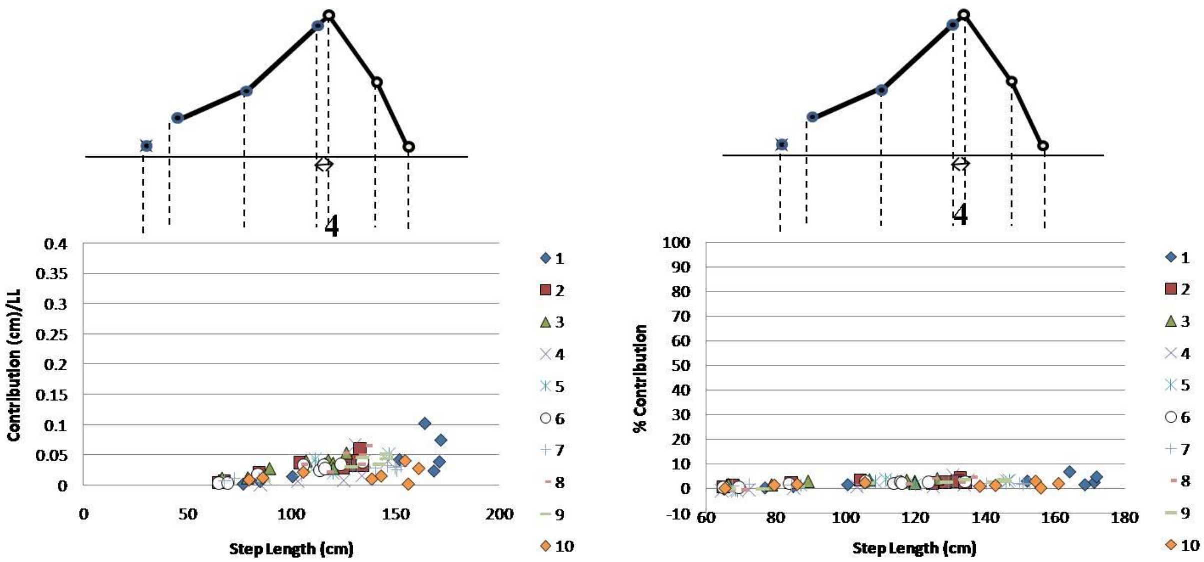

Figure 3.11: Pelvic segment contribution to step length versus actual step length for all ten subjects and all walking conditions. Contribution is shown normalized by leg length (left) and as a percentage of overall step length (right).

Contribution to step length by the leading thigh segment was similar to that of the trailing shank segment, increasing fairly linearly from 0.1 to 0.35 LL for increasing step lengths between 0.65 and 1.80 m (Figure 3.12). Leading thigh percentage contribution to step length remained around 20% for all conditions. There was a moderate, positive correlation in the relationship between step length and percent contribution of the leading thigh segment to step length ($r=0.31$, $n=90$, $p=.003$) with longer step lengths associated with higher percentage contribution by the leading thigh segment. The ANOVA found that percentage contribution to step length by the leading thigh segment was significantly different between walking conditions 1 through 5, $F(1.43,12.84)=6.85$, $p=0.015$ with Greenhouse-Geisser correction, though pairwise comparisons found no significant differences between any of the walking conditions.

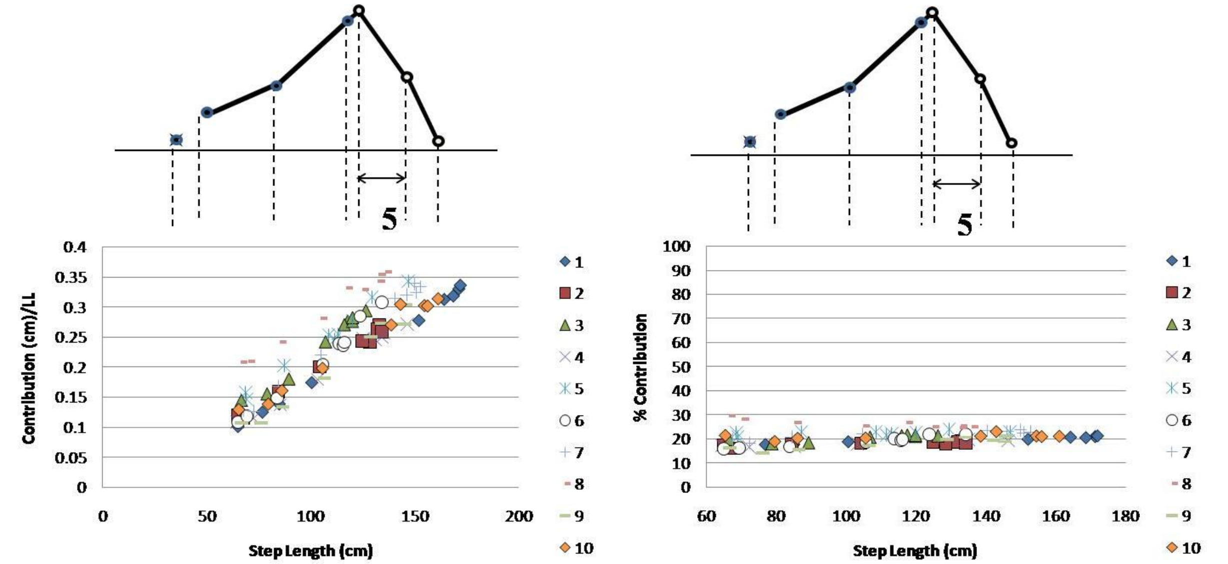

Figure 3.12: Leading thigh segment contribution to step length versus actual step length for all ten subjects and all walking conditions. Contribution is shown normalized by leg length (left) and as a percentage of overall step length (right).

The leading shank segment was the largest contributor (more than 0.15 LL) to step length for shorter step (less than 1.09 m), but as step length increased, the contribution leveled off around 0.25 LL (Figure 3.13). Therefore, the percentage contribution actually decreased from 30% to less than 20% as step length increased from 0.65 to 1.80 m. There was a strong, negative correlation for the relationship between step length and percent contribution of the leading shank segment to step length (r=-.89, n=90, P<.0005) with longer step lengths associated with lower percent contribution. One-way ANOVA determined that the percent contribution to step length by the leading shank segment was significantly different between walking conditions 1 through 5, F(1.52,13.72)=81.08, P<0.001 with Greenhouse-Geisser correction. Pairwise comparisons found that significant differences (P<0.05) occurred between all walking conditions 1 through 5.

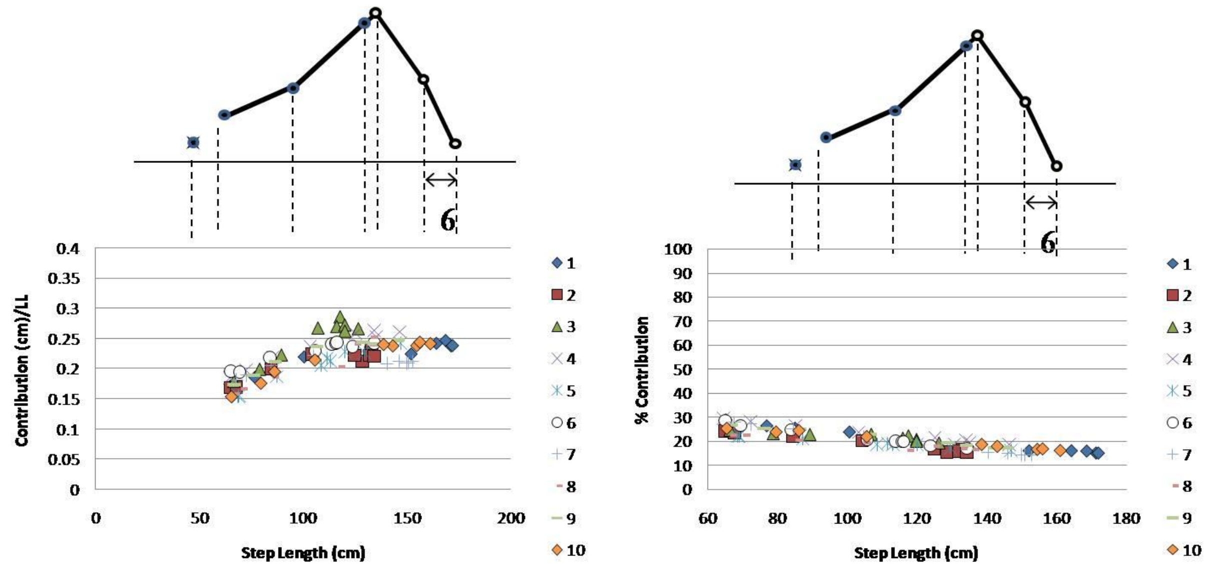

Figure 3.13: Leading shank segment contribution to step length versus actual step length for all ten subjects and all walking conditions. Contribution is shown normalized by leg length (left) and as a percentage of overall step length (right).

To better understand how each segment contributes as a percentage of overall step length, the mean values for all 10 subjects were plotted for each walking condition in Figure 3.14 and Figure 3.15. When comparing walking conditions 1 through 6 (Figure 3.14), the percent contribution of the trailing ankle-foot, trailing shank, pelvis, and leading thigh increased by 3-4% as step length increased from shortest to longest step lengths. A decrease of 3% by the trailing thigh and 9% by the leading shank occurred from shortest to longest step length.

When comparing those walking conditions that were to test training techniques of increasing step length (walking conditions 5-9), very little change was observed (Figure 3.15). During the coaching trials (conditions 7 and 8) the percent contributions of the

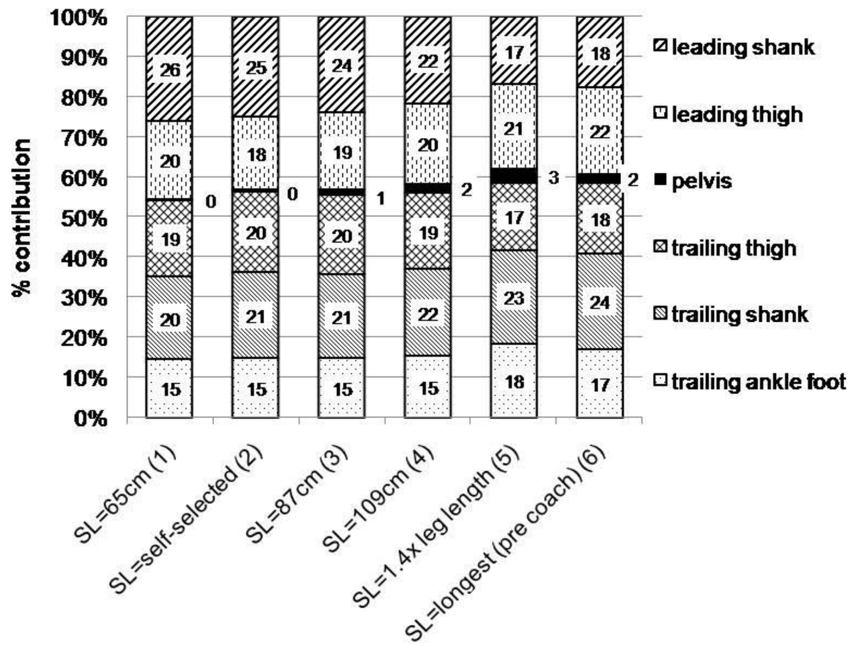

Figure 3.14: Mean percent contribution for ten able-bodied subjects of each segment to overall step length for walking conditions 1-6 (increasing step lengths).

segments that were expected to increase (trailing ankle-foot and shank for condition 7 and pelvis for condition 8) did display the highest percent contributions compared to those for the other walking conditions, but this was only by 1%. After the training techniques were performed, there did not seem to be any notable change in step length or in segment percent contributions (condition 9).

3.2.4 Ankle-foot roll over shape and Effective Foot Length Ratio

Mean ankle-foot roll over shapes were calculated for all subjects at each of the nine walking conditions. Representative roll over shapes are plotted in Figure 3.16 for the increasing step length conditions (1-6), and for longest step length, pre and post coaching techniques (conditions 5-9). Little to no change in ankle-foot roll over shapes between walking conditions 1-4 (step lengths between 0.65 m and 1.09 m) was observed. As step lengths increased (longer than 1.09 m), the roll over shape orientation changed (became increasingly "dorsiflexed") and a sharp downward movement at the end of the shape was observed, indicative of rapid plantarflexion prior to ipsilateral toe-off.

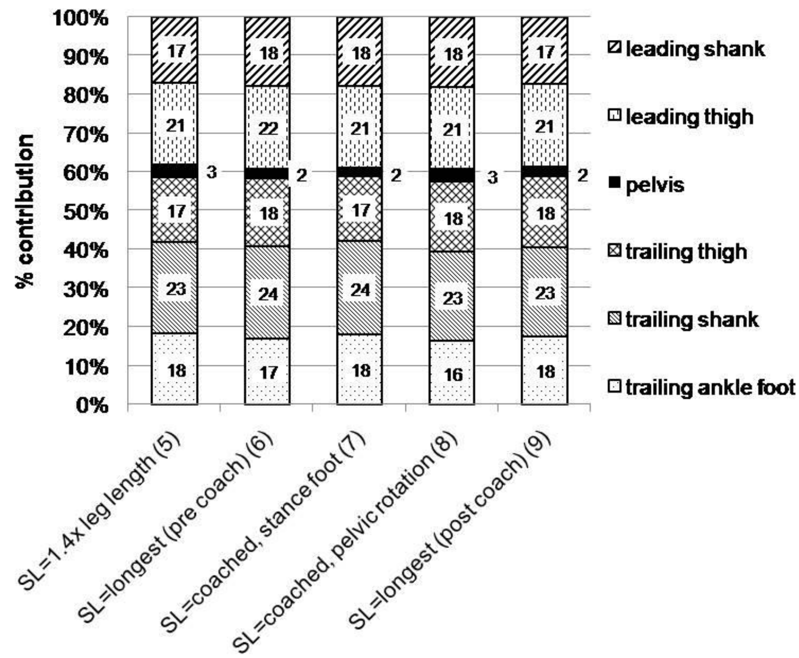

Figure 3.15: Mean percent contribution for ten able-bodied subjects of each segment to overall step length for walking conditions 5-9 (pre and post coaching conditions).

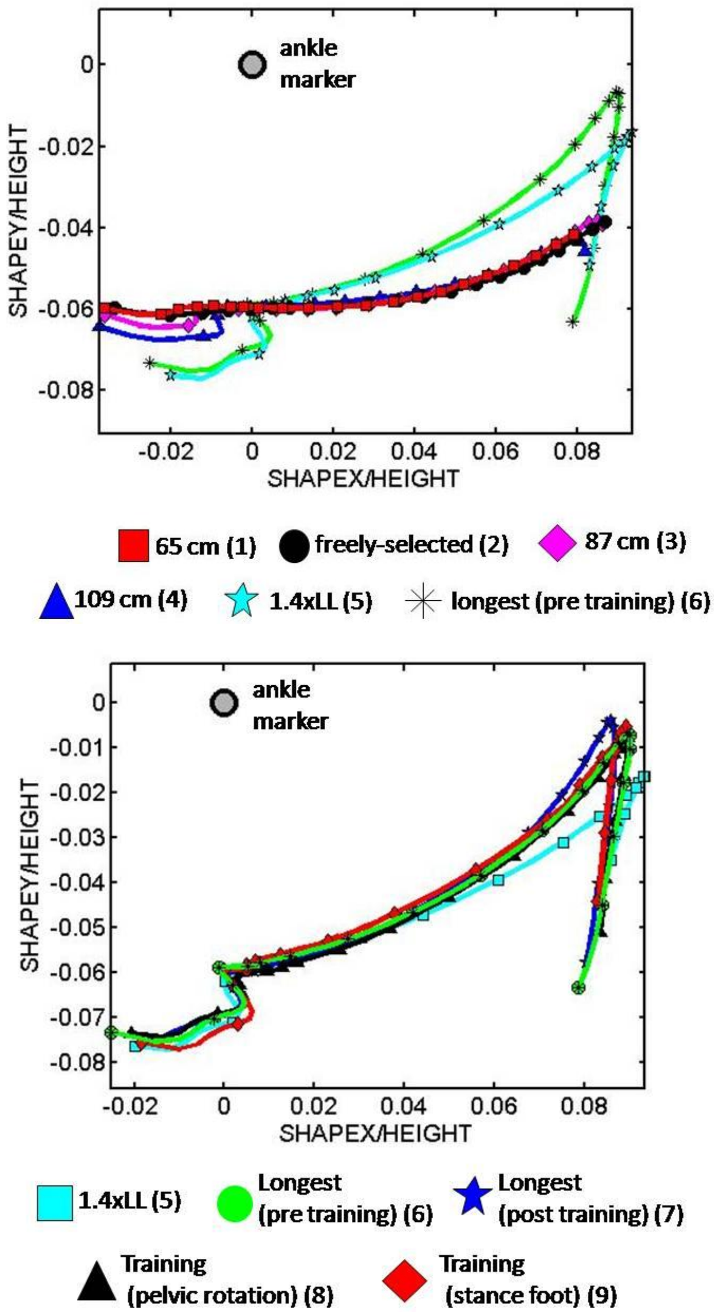

Figure 3.16: Representative ankle-foot roll-over shapes from heel contact to opposite heel contact of one subject for increasing step length walking conditions 1-6 (top) and for longest step length walking conditions 5-9, pre and post training (bottom). Little differences in ankle-foot roll over shape for pre or post training walking conditions were observed. For the long step lengths, roll over shapes are "dorsiflexed" compared to those for freely-selected walking and display downward movement at the end of the shape.

The Effective Foot Length Ratio (EFLR) was calculated for each subject at each of the 9 different walking conditions and is plotted against the measured step length in Figure 3.17. The EFLR of the physiologic foot has been reported to be 0.83 (Hansen, Childress et al. 2004). The measured mean EFLR for the 10 subjects in this study was determined to be 0.77 for freely-selected walking. For this study, as step length increased from shortest to longest step length, mean effective foot length ratio (EFLR) increased slightly, from approximately 0.7 to 0.8 times the total foot length. Overall range of EFLR for the 10 subjects was between 0.64 and 0.87.

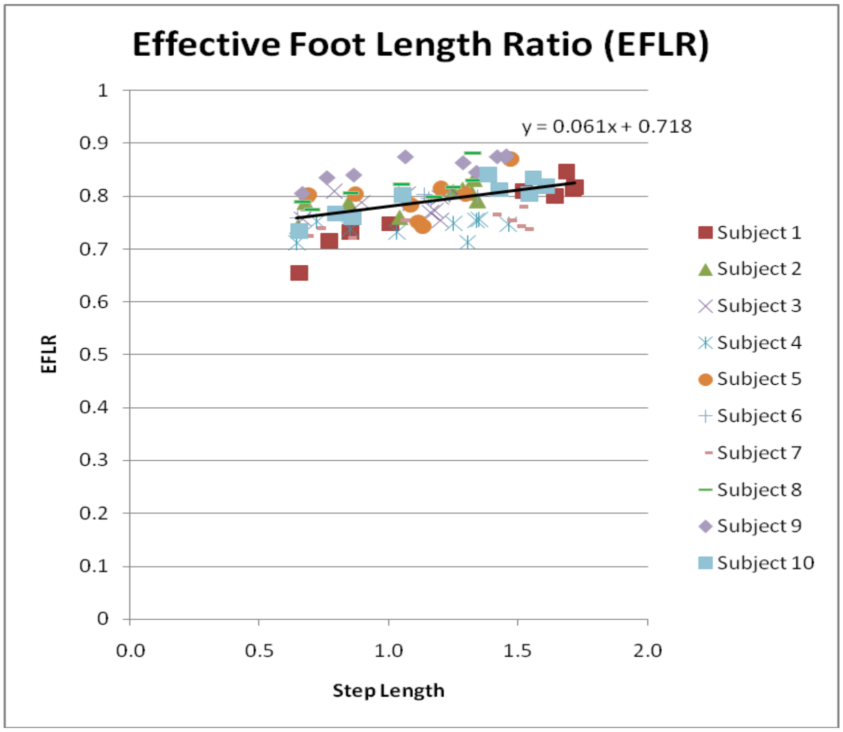

Figure 3.17: Effective Foot Length Ratio (EFLR) for all 10 subjects over a range of step lengths. EFLR is a measure of the effective foot length divided by the total foot length.

3.2.5 Discussion

Able bodied subjects displayed similar patterns of segment contributions for increasing step length amongst each other. The contributions as a percent of overall step length were also very similar over a range of different step lengths. For freely-selected walking, mean percent contribution to overall step length is highest for the shank segments, followed by the thigh segments, trailing ankle-foot segment, and then the pelvic segment. As step length increased, all segment contributions increased, with the trailing shank and leading thigh contributing the largest normalized amount to step length. This also seems to be the conclusion of Murray et al. (1966), though the results from that study were based on joint angle analyses. As a percentage contribution, the trailing thigh and leading shank decreased for increasing step length while contributions by the trailing ankle-foot, trailing shank, pelvic segment, and leading thigh increased. There is no single segment that seems to play the leading role for step length increases, as all contributions, except by the pelvis, were around 20% of the overall step length.

It has been previously stated that the pelvis plays an influential role for increasing step length (Inman et al. 1981; Murray et al. 1966; Ohmichi and Miyashita 1983; Perry 1992). Pelvic rotation range of motion measured in this study was similar to that of other studies where step length was varied. Even though pelvic rotation increased considerably (ROM increased from 8° to 40°) for increases in step length, the SCSL analysis reveals that the pelvic segment contributes very little to overall step length (maximum contribution of 3%, or around 0.05xLL). This result was somewhat surprising and is contrary to conventional thinking about gait, particularly for clinical evaluations. Even when subjects were coached to rotate their pelvis in order to take longer steps (walking condition 8), the contribution of the pelvis was still only 3% of overall step length. This small contribution is most likely due to the considerably shorter segment length of the pelvis that contributes to the forward progression of the limb ($contribution_{pelvis}=L_{pelvis}\sin\theta_{pelvis}$) compared to that of the shank and thigh.

For shorter step lengths, the leading shank segment is the largest contributor to overall step length. As step length increased, the normalized contribution by the leading shank segment also increased, but as a percentage of overall step length it actually decreased (approximately a 9% difference between shortest and longest step lengths). Percent contribution by the trailing thigh segment also decreased (4%) as step length increased. This suggests there are some limiting factors to step length by these segments. This is similar to that reported by Inman et al. (1981), who stated that even though hip flexion and extension increased with stride length, the increases tended be to greater for flexion than extension due to the anatomical constraints on these movements. Limits to muscle and tendon stretching plays a role in the limits to achievable step length (Danion et al. 2003).

The longest step length was observed for walking condition 5 (step length condition of 1.4 times leg length), with an observed mean step length of 1.42 times leg length. The trailing shank segment had the largest mean normalized contribution (0.33 LL), followed by the leading thigh segment (0.30 LL), trailing ankle-foot segment (0.26 LL), leading thigh and shank segments (both 0.24 LL), and pelvic segment (0.05 LL). Step length for this walking condition was probably longer than walking conditions 6-9 because markers on the ground served as “goal” markers that subjects were trying to reach. Conclusions from this study have determined that the maximum achievable step length for untrained healthy able-bodied ambulators is around 1.4 times leg length. The use of coaching techniques did not seem to have an effect on increasing achievable step length. Possible reasons for this are that the training time was not long enough for these subjects to utilize the coaching techniques or that they became fatigued during the experiment. It is also possible that the training techniques utilized were not effective for increasing step length, though these techniques are those used in race walking (Bumgardner 2004; McGovern 1998). Longer step lengths could probably be achieved with longer training times along with stretching and conditioning exercises.

Few differences in ankle-foot roll over shape were observed for step lengths below 1.09m. For step lengths greater than 1.09m, the ankle-foot roll over shape became increasingly “dorsiflexed”. The increase in ankle dorsiflexion during stance allows the body to progress further over the ankle compared to normal walking and thus contributes to the increased percentage of the trailing ankle-foot segment to step length. The long downward movement at the end of the roll over shape suggests that the center of pressure remains underneath the ball of the foot as the ankle plantarflexes quickly just prior to opposite heel contact. The downward “push” in the roll over shape seems to be an active component of the ankle, propelling the foot and body forward during late stance. Though the roll over shape arc length/EFLR did not change substantially, an increase in the contribution of the trailing ankle-foot and shank segments seems to substantiate an increase in heel rise at the end of stance and an active "push" by the trailing limb. This also allows a longer time in the gait cycle for the leading leg to swing out before contacting the ground.

3.3 Conclusions

These analyses show how the lower limb segments contribute to the overall step length as step length was varied. No single segment accounts for the ability to increase step length. A surprising result is that pelvic rotation does not contribute as much to step length as is generally believed. Knowing how able-bodied persons increase their step length may be useful to help in training of persons with disabilities in gait. By identifying the components of the step length that are different or deficient, targeted physical therapy or strength training may be able to help improve that segment's contribution to gait. The SCSL analysis could then be used as an outcome measure to determine if training or therapy was effective. If a prosthetic or orthotic device needs to be worn, new designs for these devices could be made to help improve upon the step length and speed, and make gait more comparable to that of able-bodied persons.

Chapter 4: Step length modulation of race walkers

4.1 Introduction

Stemming from the racing of footmen in the late 16th century, and becoming a British competitive sport called Pedestrianism, race walking was first established as a track and field event in the 1880's. It became an official Olympic event in 1908 (Howell 1996; Wallace 1989). Though the sport is one of endurance, race walkers can generally walk at much faster speeds than that of normal freely-selected walking. Teresa Vaill, a U.S. ranked women's race walker, can walk at speeds between 3.4 and 3.7 m/s for 10 and 20 km races (USA Track & Field 2008). Jonathan Matthews, a U.S. ranked men's race walker, is able to walk at speeds around 2.9 m/s for long 50 km races but can average 3.9 m/s for shorter 20 km races. The female and male records for 20 km races are held by Olimpiada Ivanova and Bernardo Segura, whose record speeds were 3.8 m/s and 4.3 m/s, respectively (USA Track & Field 2008). In contrast, freely-selected walking speeds of healthy able-bodied ambulators are between 1.0 and 1.4 m/s, while a very fast walking speed measured in our laboratory is around 2.2 m/s. Besides increases in cadence, race walkers are able to take longer step lengths than untrained persons to achieve these higher speeds (Murray et al. 1983). The discrepancy between speeds of proficient race walkers and untrained able-bodied persons implies there may be training methods that can increase the step length (and speed) of walking.

According to the USA Track and Field (USATF) rules, "race walking is a progression of steps so taken that the walker makes contact with the ground so that no visible (to the human eye) loss of contact occurs” (USA Track & Field 2008). Race walking rules also require that the advancing limb remain straightened (no knee flexion) from the time of initial foot contact with the ground until the leg is in the vertical upright position (Figure 4.1) or when the body has advanced over this limb. This is unlike that of normal walking, where stance phase knee flexion is observed. This stance phase knee flexion increases with increasing step length for self-selected walking (Murray et al. 1966). Other differences in gait are those believed to enhance race walking performance. It has been suggested by race walking trainers that to increase walking speeds, the rear foot should stay on the ground as long as possible and provide a powerful push-off at the end of stance as the foot rolls up onto the toes. Pelvic rotation should also occur so that the feet walk along a straight line in front of the body (Bumgardner 2004; McGovern 1998; 2005).

Figure 4.1: Photographs of race walking subject from times between heel contact and toe-off. The photos show how the leg stays straight until the body advances over the leg (adapted from Salvage, 2005).

4.2 Previous research

A few studies have performed basic gait analyses on race walkers. Hoga et al. (2003) videotaped 28 elite male race walkers participating in official 20 km races, and created a 14-segment model to calculate biomechanical parameters from the videos. Results indicated that walking speed was significantly related to the step length rather than step frequency. They also concluded that height plays a factor in the speed of race walkers. An analysis of the lower limb kinematics was performed by Cairns et al. (1986), who showed that race walkers had significantly increased dorsiflexion of the ankle just prior to heel strike, knee hyperextension during the stance phase of gait, increased hip flexion during the swing phase of gait, and greater overall pelvic tilt, rotation, and obliquity compared to normal walking.

Murray et al. (1983) had similar findings when comparing two Olympic race walkers with normal men walking at fast speeds. Speeds of approximately 3.3 m/s were measured in the race walkers compared to 2.4 m/s of the normal men walking at their fastest speeds. Race walkers exhibited less ankle dorsiflexion during stance and more ankle dorsiflexion at end of swing phase. Knee hyperextension of 8° more and pelvic rotation of 20° more than that of the normal group were also measured. Higher hip flexion peaks were observed during swing, but race walkers immediately reversed into hip extension prior to initial contact. In contrast, peak hip flexion for normal fast walking occurred during the late swing phase of gait and was sustained through early stance. A more gradual reversal from hip flexion to extension was observed. Along with kinematic data, Murray et al. also obtained EMG of the subjects and noticed an increase in the amplitude and duration of the EMG signal in the limbs of the race walkers compared to the normal subjects walking fast.

The findings of the study described in Chapter 3 of this dissertation reported that to increase the step lengths of able-bodied persons, an increase in contributions of all the lower limb segments (trailing ankle-foot, shank, and thigh segments, pelvis and leading shank and thigh segments) occurred. The contribution by the pelvis was considerably less than that of the other segments as a percentage of overall step length. The walking speeds observed in research subjects were still considerably less than that measured for proficient race walkers, even after brief training that involved techniques used for coaching race walkers. Performing gait analyses of proficient race walkers by observing kinematics, segment contributions to step length and ankle-foot roll over shapes will allow us to better understand how race walkers are able to walk at such fast speeds. The segment contribution to step length (SCSL) analysis will also determine how race walkers differ in their methods to increase their step length and might be used as a tool to determine where improvements in walking performance can be made.

4.3 Methods

4.3.1 Gait data acquisition

Eight able-bodied subjects who stated they were competitive race walkers signed consent forms that were approved by Northwestern University's Institutional Review

Board. Subjects wore athletic shoes and comfortable clothing during the study. Data collection and analyses for the study were conducted in the VA Chicago Motion Analysis Research Laboratory (VACMARL). An eight-camera Eagle Digital Real-Time motion measurement system (Motion Analysis Corporation, Santa Rosa, CA) was used to acquire marker movements at 120 Hz and calculate kinematic data. Ground reaction forces were acquired using six AMTI (Advanced Mechanical Technology, Inc., Watertown, MA) force platforms simultaneously recorded with the motion analysis cameras at 960 Hz. A modified Helen Hayes marker set (Kadaba et al. 1990) was used to define a biomechanic model on each person. A static standing trial was performed before the walking experiment in order to estimate the location of the joint centers of rotation.

Subjects were asked to walk at a variety of walking speeds. Data for this study were collected with the subjects walking at their freely-selected walking speed and then very fast walking speed. They were then asked to walk using their race walking technique. Walking trials were repeated until 3-5 clean force platform hits were obtained for each leg.

4.3.2 **Data analyses**

Data were processed using Motion Analysis' EVa and Orthotrak software. Missing data points in marker position data were interpolated using a cubic spline technique. Raw marker position data were filtered using a fourth-order bidirectional Butterworth infinite-impulse response digital filter with an effective cutoff frequency of 6.0Hz. Data were

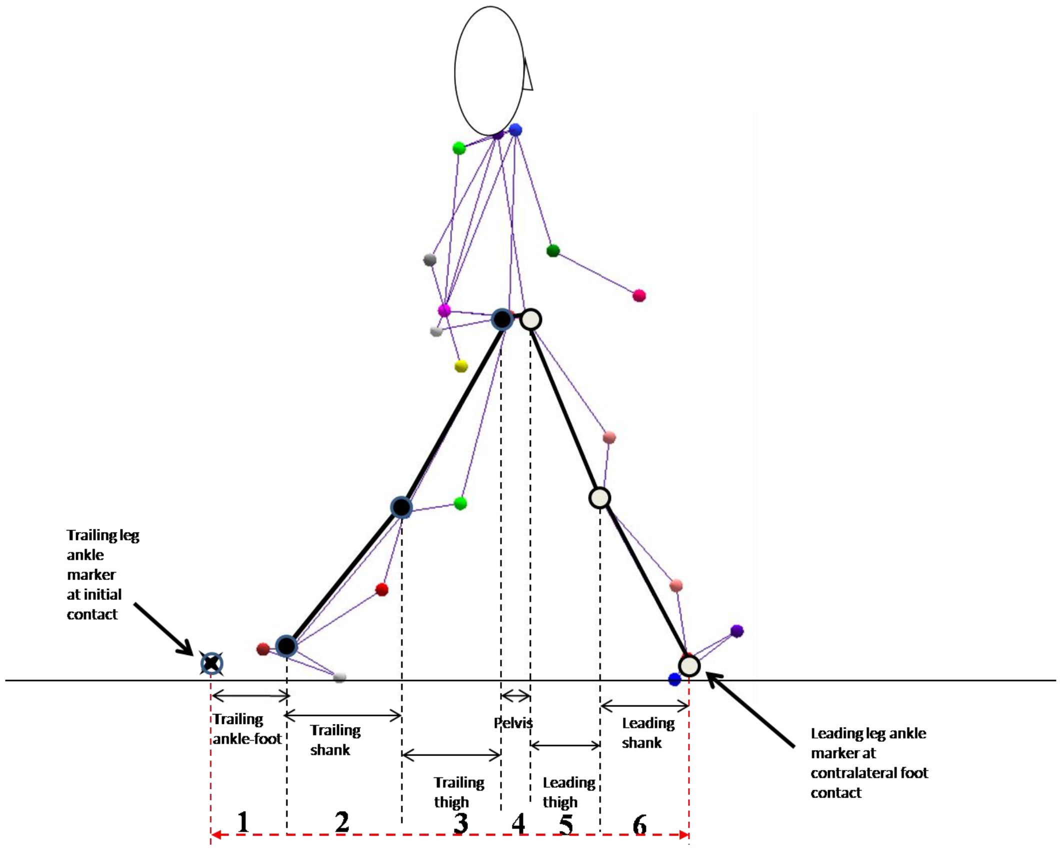

Figure 4.2: Representation of segments used for the Segment Contribution to Step Length analysis. Segment 1, the trailing ankle-foot segment, is the sagittal distance traversed by the trailing ankle joint center from foot contact to opposite foot contact. Segments 2-6 (shank, thigh and pelvis segments) are the measured lengths of these segments at the time of initial contact of the leading limb. The lengths of each shank segment is measured from ankle joint center to knee joint center, while the thigh segments are measured from knee joint center to hip joint center. The pelvic segment length is measured as the distance between the two hip joint centers.

further processed using custom programs in Microsoft Excel (Microsoft Corporation, Redmond, WA) and Matlab (The Mathworks, Inc, Natick, MA). It was assumed that subjects walked fairly symmetrically on both sides. Data for step length analysis were calculated from left side steps.

Specific gait data that were analyzed for this study were walking speed, step length, cadence, and double limb support time for all walking conditions as well as sagittal plane ankle, knee, and hip flexion, and transverse pelvic rotation kinematics. A Segment Contribution to Step Length (SCSL) analysis was also performed. The SCSL analysis calculates the contribution of six lower limb segments to the overall step length. The segments that were included were the trailing ankle-foot, shank, and thigh segments, the pelvic segment, and the leading thigh and shank segments (Figure 4.2). Specifics of how each segment contribution is calculated are provided in Chapter 2. Data were normalized by leg length to eliminate effects of longer leg lengths of some subjects. Segment contributions were also reported as a percent contribution to overall step length. Gait analyses also included ankle-foot roll over shape characteristics and effective foot length ratios (EFLR). Roll over shapes are created by transforming center of pressure data from the laboratory-based coordinate system to a body-based coordinate system (Hansen, Childress et al. 2004). The EFLR is a measure of the distance the COP progresses under the foot and is calculated as the length of the roll over foot shape (i.e. the distance from the heel to the anterior end of the shape) divided by the total foot length (Hansen, Sam et al. 2004).

Temporospatial and kinematic data for the freely-selected walking condition were compared to that from ten able-bodied subjects with no race walking training. Data from these subjects were collected in our laboratory previously (mean age, weight, and height was 25 ± 1 years, 69.3 ± 9.5 kg, and 174.6 ± 9.1 cm).

4.4 Results

4.4.1 Subject info and temporal spatial data

Table 4.1: Subject's Vital Statistics

Subject	Gender	Age	Mass (kg)	Height (cm)	Leg length (cm)
1	M	22	66.0	171.5	95.25
2	F	58	51.0	162.0	96.08
3	M	26	75.5	189.0	106.81
4	M	26	71.5	171.0	106.03
5	M	34	80.0	168.5	94.11
6	F	24	58.0	158.0	90.49
Mean (Standard Dev)	**-**	**31.7 (13.5)**	**67.0 (11.0)**	**169.6 (11.9)**	**98.13 (6.71)**

Of the 8 race walking subjects that were tested, only 6 subjects displayed proper race walking techniques (no stance phase knee flexion until the leg was in the vertical upright position and one foot on the ground at all times). Data reported for this study is from these 6 race-walking subjects. Vital statistics are reported in Table 4.1. Average age, mass, and height were 32 years, 67.0 kg, and 169.6 cm, respectively. Mass and height of the race walking group were similar to that of the able-bodied subjects, though the normal subjects' mean age was 7 years younger than the race walking group. Table 4.2 lists the mean temporospatial measurements for the race walkers as well as from the 10 able-bodied walkers that were previously acquired. The freely-selected walking speed of the race walkers was 0.14 m/s faster than that of the able-bodied persons with no race walking training, though

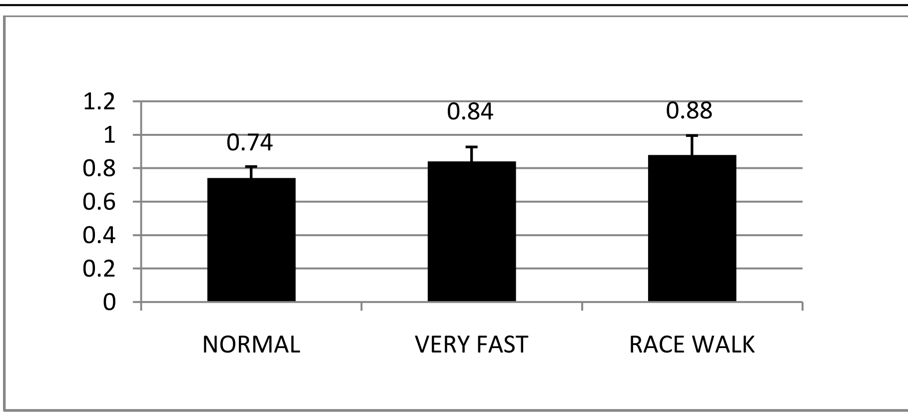

Figure 4.3: Mean step length (normalized by leg length) for 6 subjects walking at self-selected normal, very fast, and race walking conditions. Vertical error bars represent the standard deviation for the group.

mean step lengths were similar. Cadence for freely-selected walking on average was 14 steps/min more for the race walkers than that for the untrained subjects. As the race walkers switched from normal walking to very fast walking and then to race walking, mean speed increased by 0.62 m/s and 1.11 m/s, respectively. Speed changes were due to increases in both step length and cadence. Step length was also normalized by leg length (Figure 4.3) for the race walkers' data, which increased by 0.14 times leg length when comparing normal and race walking conditions.

Walking Condition	Mean Speed (m/sec)	Mean Step Length (m)	Mean cadence (steps/min)	Double support time (as % of gait cycle)
Normal (untrained subjects)	1.35 ± 0.16	0.73 ± 0.04	110 ± 8	11 ±1%
Normal (race walkers)	1.49 ± 0.19	0.72 ± 0.06	124 ± 12	11 ± 1%
Very fast (race walkers)	2.11 ± 0.22	0.82 ± 0.08	157 ± 23	9 ± 1%
Race walk (race walkers)	2.60 ± 0.37	0.87 ± 0.12	181 ± 18	5 ± 3%

Table 4.2: Walking condition and respective mean speed, step length, cadence, and opposite toe-off time for the 6 race walkers. Gait was symmetrical, so contralateral initial contact occurred at 50% of the gait cycle.

4.4.2 **Kinematics**

Mean sagittal plane ankle, knee, and hip angles, as well as pelvic rotation in the transverse plane over the gait cycle during freely-selected walking of both groups are plotted in Figure 4.4. Kinematics at the ankle, knee, and hip were similar between the race walkers and non-trained able-bodied walkers. An increase of 7° of peak to peak pelvic rotation was measured in the race walkers. A phase shift was also observed, with pelvic rotation peaks occurring during initial stance (<10% of the gait cycle) for the race walkers and more towards midstance (approximately 25% of the GC) for the able-bodied subjects.

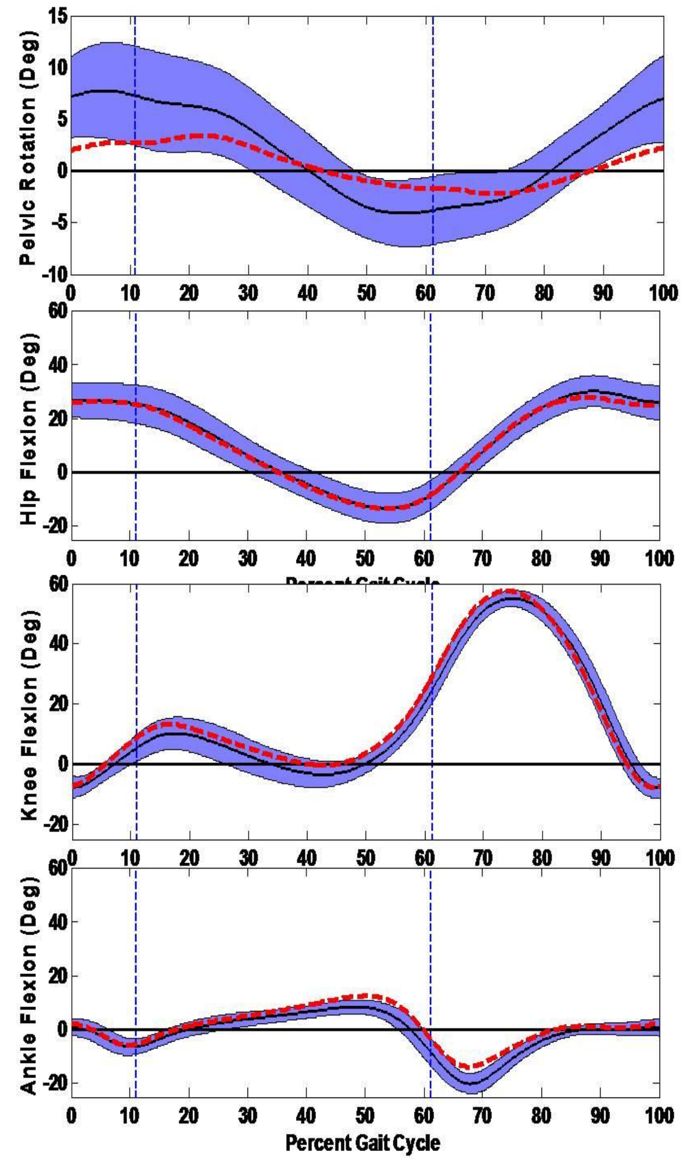

Figure 4.4: Mean sagittal ankle, knee, and hip flexion, and pelvic rotation kinematic curves for six race walking subjects walking at freely-selected walking speeds (solid black line). Shaded region represents one standard deviation of the subjects. The dashed lines are from mean kinematic data of ten (untrained) able-bodied subjects walking at freely-selected walking speeds.

Comparisons between the three walking conditions of the race walking group were also performed (Figure 4.5). An increase in pelvic rotation range of motion (ROM) was observed between the different walking conditions (14°, 17°, and 19° for normal, fast, and race walking, respectively). Between the normal and race walking conditions, pelvic rotation ROM increased by 5°.

An increase in mean hip flexion range of motion from 43° to 48° was observed between normal and race walking conditions. Though fast walking hip flexion ROM (47°) was similar to that for race walking, hip flexion peak during race walking was higher

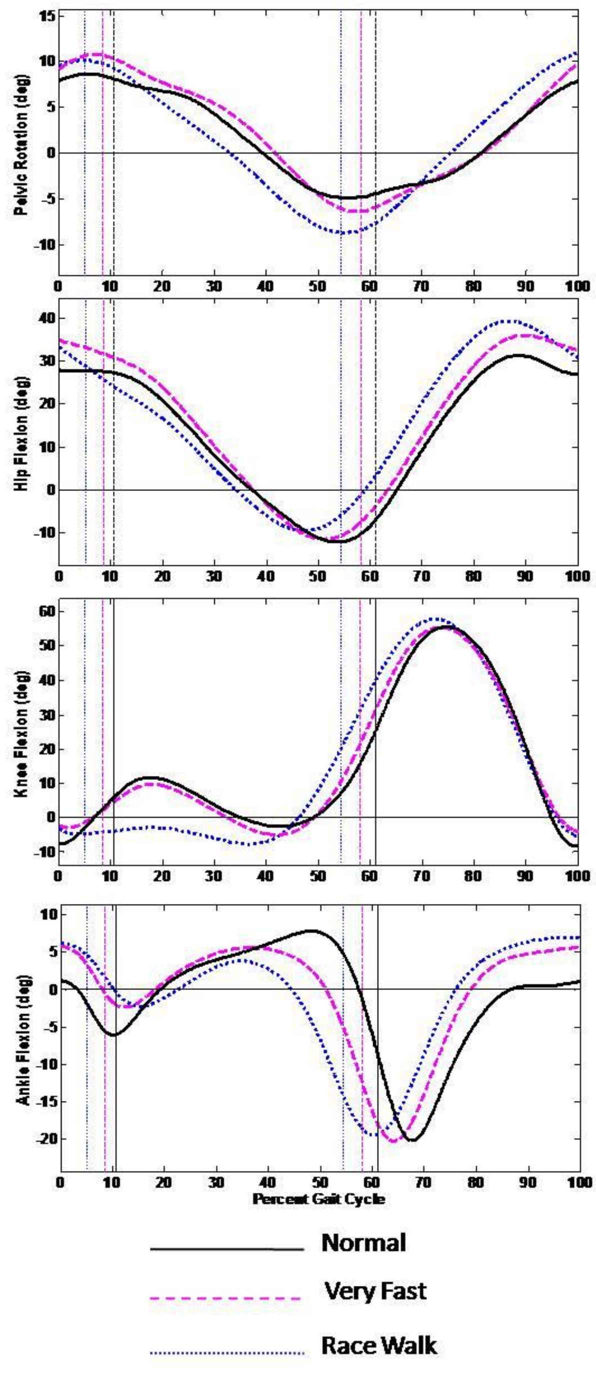

Figure 4.5: Mean sagittal plane ankle, knee, and hip, and transverse plane pelvic rotation kinematic curves for six race walkers during normal (freely-selected), very fast, and race walking conditions. Vertical lines depict times of contralateral toe-off (between 5 and 11% of the gait cycle) and ipsilateral toe-off (between 55 and 61% of the gait cycle).

compared to fast walking (39° vs 36°) and hip extension peak was lower for race walking compared to fast walking (9° vs 11°). At initial contact, hip flexion angles for fast and race walking conditions were similar (33°), which was approximately 5° higher than that measured during normal walking. Hip flexion peaks for the three walking conditions occurred around 85% of the gait cycle, with flexion decreasing as the swing leg began extending at the end of the gait cycle.

Subjects displayed knee hyperextension at initial contact for all walking conditions, though it was actually highest for the freely-selected walking condition at this time (-8°). Besides the differences in knee hyperextension at the beginning of the gait cycle, kinematic knee flexion curves were similar between normal and fast walking. While the subjects were race walking, the knee remained in hyperextension until 40% of the gait cycle, so no stance phase knee flexion occurred. Knee flexion peak during swing was highest for race walking.

For very fast and race walking conditions, the ankle joint was dorsiflexed by approximately 6° at initial foot contact, and only plantarflexed by 2° during stance. In comparison, the ankle was in 1° dorsiflexion at initial contact for normal walking and plantarflexed by 6° during stance. For increasing walking speeds, the ankle joint dorsiflexed earlier in the gait cycle, before opposite foot contact occurred.

4.4.3 Segment Contribution to Step Length analysis

Segment contributions to step length of the six lower limb segments were also calculated for the race walking group. Data were reported as a value normalized by leg length (Figure 4.6), and also as a percentage of the overall step length (Figure 4.7). As walking speed increased, actual contribution to step length by the trailing ankle-foot and shank segments increased by 0.04 LL. Only slight changes were observed in the other segments. Contributions by the leading and trailing thigh segments were highest for the very fast walking condition compared to normal and race walking.

As a percent contribution to overall step length, the trailing ankle-foot and shank segment as well as the pelvic segment increased for increasing step length, while the trailing thigh, and leading shank and thigh segment contributions decreased for increasing step length. Largest changes were observed by the trailing ankle-foot segment, with a 4% increase in contribution between normal and race walking, and by the leading shank segment, with a 5% decrease in contribution between normal and race walking. The other segment contributions changed by 2% or less. The lower limb segments of a representative race walker were plotted to illustrate how the segments contribute to the overall step length for freely-selected and race walking (Figure 4.8). As can be observed, the trailing ankle-foot "rolls" further forward, and an increase in the heel rise contributes to the forward progression of the trailing limb. Few differences

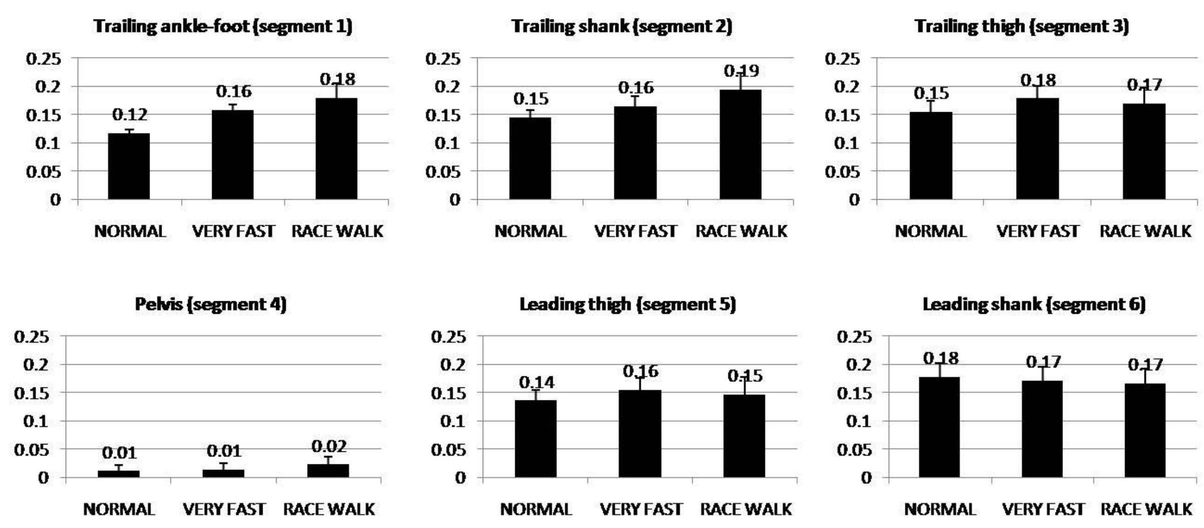

Figure 4.6: Mean contribution of the six lower limb segments used in the SCSL analysis to overall step length (normalized by leg length) for freely-selected (normal), very fast, and race walking speeds of the six subjects. Vertical error bars represent the standard deviation for the group.

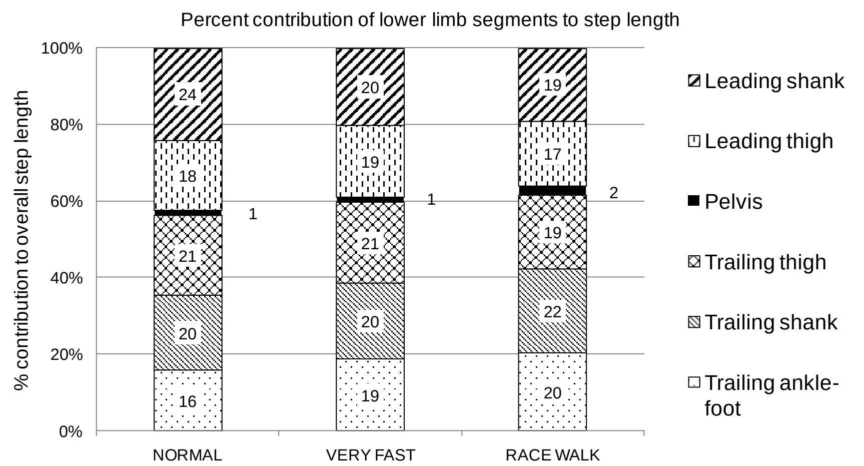

Figure 4.7: Mean percent contribution of the six lower limb segments to overall step length for the six race walkers at three different walking conditions.

were observed in the pelvis and leading limb.

Comparisons of the percent contributions of each segment were also made between the six subjects while race walking (Figure 4.9). Most of the subjects' segmental contributions were similar to each other (i.e. near the mean value), so they followed a similar pattern of movement. Subject 4, however, had lower contributions to overall step length by the trailing ankle-foot and shank, but had larger percent contributions by the thigh segments and the leading shank segment compared to the average data.

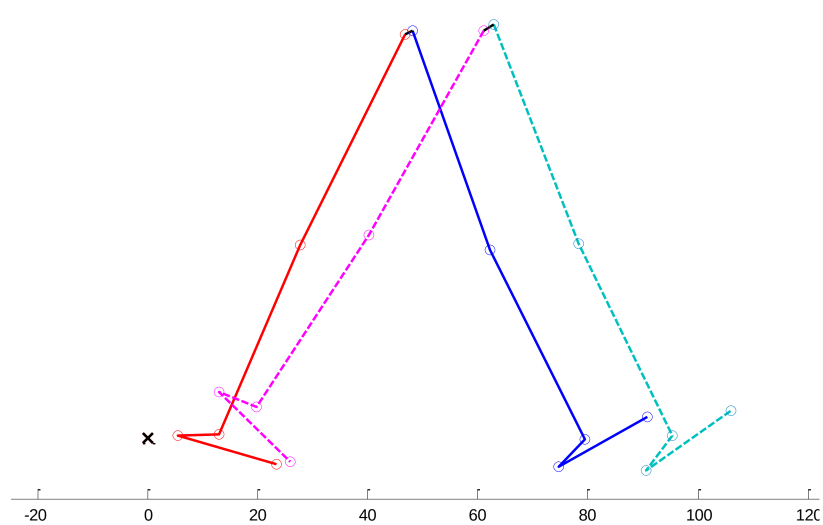

Figure 4.8: Lower limb stick figure of a representative race walker taking a step during self-selected walking speed (solid lines) and while race walking (dotted lines). X marks the position of the trailing ankle joint center at the beginning of the step for both walking conditions. For this subject, the difference in step length between the two walking conditions is approximately 15 cm (x-axis is in cm).

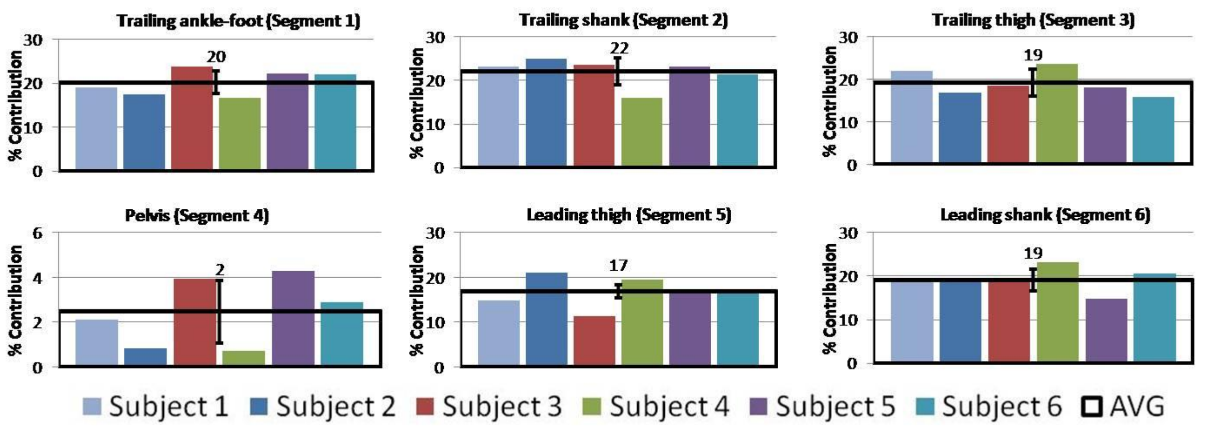

Figure 4.9: Lower limb segment contributions to step length for six subjects while race walking. Mean value of all subjects is represented by the black horizontal line across the graph. Standard deviation is plotted as the vertical line between Subjects 3 and 4.

4.4.4 Ankle-foot roll over shape and Effective Foot Length Ratio

Mean ankle-foot roll over shapes were calculated for the race walking subjects at the three walking conditions from the time of ipsilateral initial contact to contralateral initial contact. Representative roll over shapes are plotted in Figure 4.10. Mean effective foot length ratios (EFLR) of the 6 subjects were also calculated (Figure 4.11). The roll over shapes for normal and very fast walking conditions were similar, as were EFLR values (0.73 for normal walking and 0.74 for fast walking). The roll over shapes during race walking had similar shapes to normal and fast walking through most of stance phase. Towards the end of stance, a downward “push” by the foot was observed and was associated with the ankle plantarflexion observed prior to contralateral initial contact. Mean EFLR for ankle-foot roll over shapes during race walking was 0.77, an increase of 0.03 compared to normal walking of the race walkers.

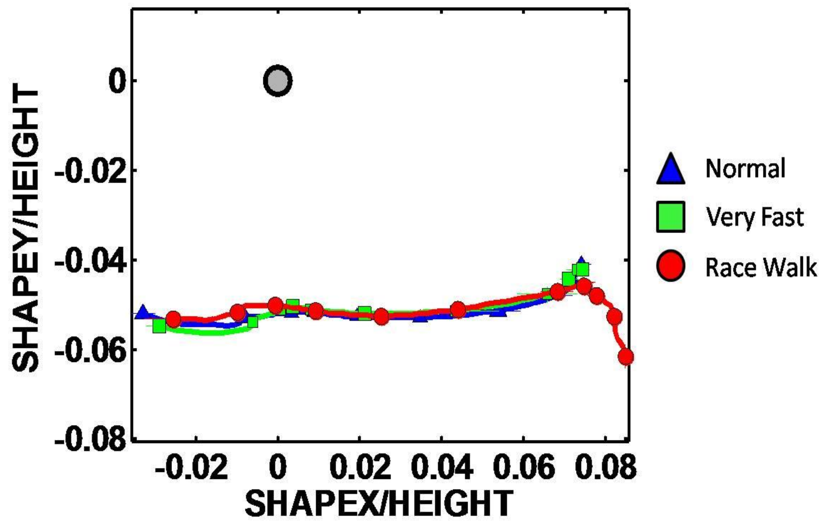

Figure 4.10: Representative ankle-foot roll over shape for subject walking at the three walking conditions. The shape is calculated from the time of initial foot contact to the time of opposite foot contact. Normal and very fast walking conditions have very similar roll over shapes. The roll over shape for race walking displays a downward movement at the anterior end of the shape, reflecting rapid plantarflexion at the ankle.

4.5 Discussion

The kinematic results of this study are in agreement with those of Cairns et al. (1986) and Murray et al. (1983). In particular, the results indicate that race walkers have increased ankle dorsiflexion at initial contact, less ankle dorsiflexion during stance, knee hyperextension during the stance phase of gait, increased hip flexion during the swing phase of gait, and greater pelvic rotation compared to normal walking.

Race walkers ambulated at faster freely-selected speeds than non-trained able-bodied individuals. This was due to a higher cadence rather than longer step length. Sagittal plane lower limb kinematics of race walkers were similar to non-trained able-bodied walkers during freely-selected walking, though they displayed slightly greater pelvic rotation range of motion. This translated into approximately 2 cm (2.23 cm vs 0.26 cm) or a 1% larger contribution by the pelvic segment to overall step length. Though the contribution by the pelvis was still small, it was always a positive value for race walkers, whereas for non-trained able-bodied walkers, pelvic rotation sometimes displayed negative contribution to overall step length (Chapter 2).

Race walkers tended to make adjustments in the stance (i.e., trailing) leg as opposed to the swing (i.e., leading) leg when race walking compared to normal walking. There was a much larger increase in stance ankle-foot contribution to step length while few changes were observed in the leading limb. Thus, as step length increased for race walking, the percent contributions to step length of the swing thigh and shank decreased. It has been suggested that "overstriding" causes the advancing foot to strike the ground, preventing a smooth forward motion (McGovern 1998) and increasing energy cost. Coaches have thus suggested that lengthening of the step length should occur on the trailing limb, keeping the stance foot on the ground and using it for "powerful push off" through the

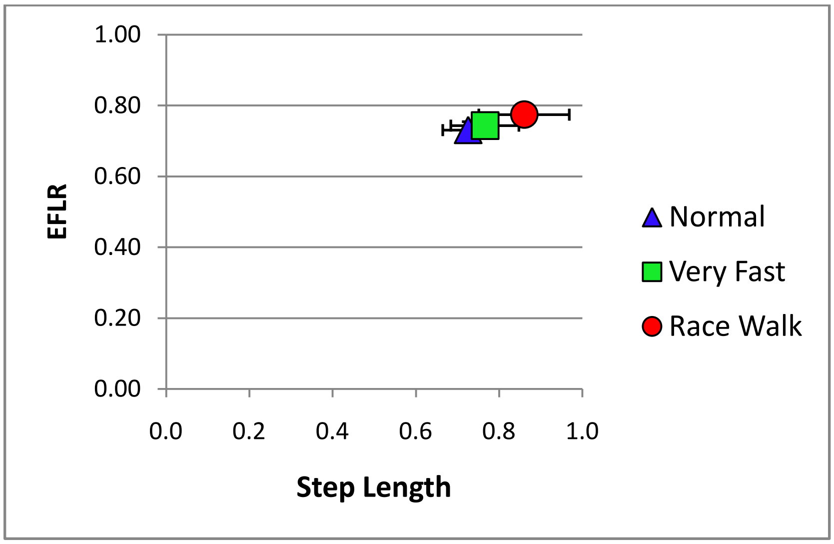

Figure 4.11: Mean Effective Foot Length Ratio for subjects walking at the three walking conditions plotted versus step length. Step length increases for increasing walking condition speed (i.e. shortest at normal speed, longest at race walking speed), and mean EFLR also increases. Vertical and horizontal error bars represent the standard deviation for the group. Standard deviation for EFLR is small (0.024, 0.034, and 0.027 for normal, very fast, and race walking, respectively), so vertical error bars may not be visible.

toes. In the race walking subjects, the earlier plantarflexion before opposite foot contact suggests this active “push-off” of the ankle joint. This phenomenon can also be observed in the ankle-foot roll over shape as a downward movement in the shape at the end of stance. While race walking, the trailing foot seems to roll forward and push-off on the forefoot, extending the ankle further forward and propelling the trailing shank segment forward. Most of the subjects seemed to walk with this pattern of gait, though Subject 4 displayed larger contributions to step length by the leading limb. The speed (2.3 m/s) and step length (0.84 m) of this subject were comparable with the other race walkers, so it is possible that there is more than one pattern of movement possible to achieve race walking technique and speeds, though energy cost may be different.

Though the ankle-foot roll over shape of the race walkers were similar for the three walking conditions through most of stance, the roll over shape was different to that observed of able-bodied persons walking at long step lengths. Particularly, the roll over shape of untrained able-bodied persons became increasingly “dorsiflexed” for long step lengths compared to normal walking as ankle dorsiflexion angles increased (Chapter 3). In contrast decreased dorsiflexion during stance was observed by the race walkers so the arc length shape remained the same until the forefoot was reached. Earlier plantarflexion activity that occurred before opposite foot contact contributed to the sharp downward movement in the anterior end of the shape. In general, the sagittal ankle kinematics suggest that the ankle motion is actively controlled to allow a smooth progression at these long step lengths and high walking speeds. An increase in EMG activity observed by Murray et al. (1983) also concurs with this active control for race walking.

Contrary to general belief (Bumgardner 2004; Inman et al. 1981; McGovern 1998; 2005; Perry 1992), pelvic rotation does not play a large role in step length increases, even for race walking, but it may be important for effective race walking by reducing the step width and, thus, possibly energy cost. This study has also shown that the leading limb does not contribute to increases in step length for race walking, and that the pelvis only contributes a small amount towards step length. Though changes in percent contribution of the trailing limb segments may seem small, only a slight increase in the percentages can have a considerable affect on overall step length. Teaching techniques of trying to keep the stance foot on the ground longer (and thus rolling and pushing off with the ball of the foot) may help to increase the step length of able-bodied walkers. The use of the SCSL and roll over foot shape in analyses are useful tools in determining if race walkers are utilizing the correct techniques and could improve upon their walking speeds.

Chapter 5: Study of roll over shape arc length for persons with bilateral trans-tibial amputation

5.1 Introduction

Previous studies have shown that persons with unilateral trans-tibial amputation walk with step length asymmetry and/or increased loading on their sound limbs compared to their prosthetic limbs (Barth et al. 1992; Isakov et al. 1997; Macfarlane et al. 1991; Powers et al. 1994; Torburn et al. 1990; Wagner et al. 1987). It is believed that the amount of asymmetry observed may be due to the type of prosthetic foot that is worn (Hansen et al. 2006; Lehmann et al. 1993; Powers et al. 1994; Snyder et al. 1995). Specifically, Hansen et al. (2006) reported that when the forefoot arc length of the effective foot rocker was reduced, a “drop-off” occurred on the prosthetic limb along with increased loading on the sound limb. As the end of the prosthetic roll over shape was reached, a rapid weight transfer from the prosthetic side onto the sound side occurred. Hansen also observed a decreased stance phase peak ankle dorsiflexion moment. Although not significant, an increase in step length asymmetry was measured when the arc length of the prosthetic foot was shortened.

By walking with prosthetic feet that have shorter than normal effective foot rockers, persons with amputation may try to compensate by deviating from normal walking patterns (e.g. increased knee flexion or pelvic rotation, or other changes in lower limb joint range of motion, etc.) that could lead to injury over time. It is possible that the sound limb is able to compensate for the changes in arc length on the prosthetic side.

By changing the forefoot arc length of the effective foot rocker on persons with *bilateral* amputation, it is possible to more directly observe differences in gait due to these changes. It is hypothesized that shortening the effective forefoot rocker arc length will result in reduced step lengths and walking speeds, and also increased 1st peak vertical ground reaction forces.

5.2 Methods

5.2.1 Data acquisition

Preliminary analysis suggested that ten subjects were needed to determine differences equivalent to one standard deviation of the measurements, assuming allowable type I error (α) of 5% and type II error (β) of 20% (statistical power of 80%) (Lieber 1990). Twelve subjects with bilateral trans-tibial amputation who walked with endoskeletal prostheses participated in the study. They signed consent forms that were approved by Northwestern University's Institutional Review Board. Data collection and analyses for the study were conducted in the VA Chicago Motion Analysis Research Laboratory (VACMARL). An eight-camera Eagle Digital Real-Time motion measurement system (Motion Analysis Corporation, Santa Rosa, CA) was used to acquire marker movements at 120 Hz. Ground reaction forces were acquired using six AMTI (Advanced Mechanical Technology, Inc., Watertown, MA) force platforms simultaneously recorded with the motion analysis cameras at 960 Hz.

A modified Helen Hayes marker set (Kadaba et al. 1990) was used to define a biomechanical model of each person. Markers for the ankle, heel, and toe were placed on a specialized plate between the pylon and the foot, at the level of the ankle. This plate was used to allow for comparison between this study and an earlier study of unilateral trans-tibial prosthesis users and to prevent variability in measurement due to changes in marker placement when removing the shoe to alter the prosthetic foot. A static standing trial was performed before the walking experiment in order to estimate the locations of the joint centers of rotation.

The protocol for the experiment was similar to that performed on persons with unilateral trans-tibial amputation by Hansen et al. (2004). Subject's prostheses were removed and taken to a separate laboratory where the sockets were disconnected from the rest of the prostheses and Shape&Roll prosthetic feet were attached to the participant's sockets using a rigid pylon. These feet are made of a copolymer plastic and are designed to conform to the roll over shape (i.e., effective rocker shape that is formed while walking) of the able-bodied ankle-foot system during walking (Sam et al. 2004). Shape&Roll feet were used for this experiment because specific changes could be implemented without altering the other components of the prostheses. The type and size of the feet were based on the height, weight, and shoe size of each participant. The length of the prostheses was held the same as that of the subject's original device. After the new components were assembled, a foam cover and stocking were placed on the foot to blind the foot type to the user, and the subject's original shoes were placed on the prosthetic feet.

The prostheses were returned to the user and a dynamic alignment was performed by a qualified prosthetist. After the subject was comfortable walking with the new prosthetic configuration, he was asked to walk at self-selected normal, slow, and fast walking speeds with the unaltered Shape&Roll prosthetic foot. This foot type was labeled as the LONG roll over shape arc length condition because it was the foot type with the longest arc length measured previously using quasi-static testing (Hansen et al. 2006) of the three feet used in this study.

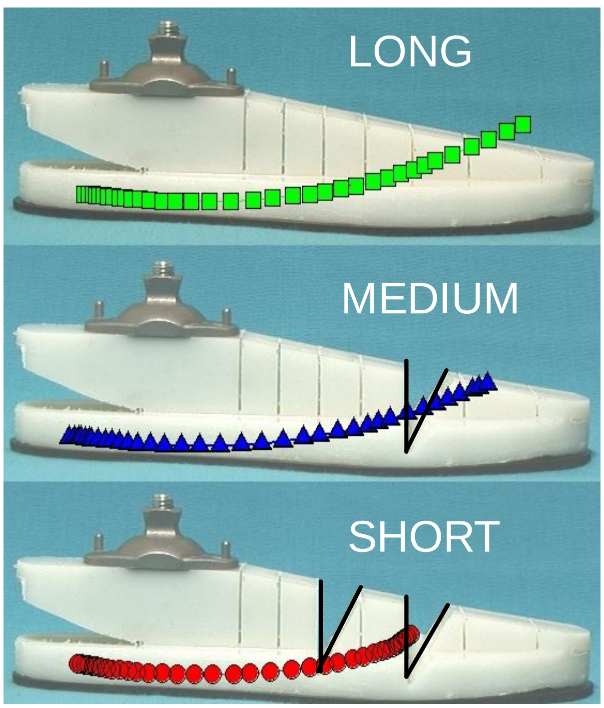

Figure 5.1: Three prosthetic foot types used in this study. The unmodified Shape&Roll foot represents the LONG foot. A wedge cut is made at 70% of the foot length for the MEDIUM foot type to shorten the effective foot length. A second wedge cut is made at 60% of the foot length for the SHORT foot type to further shorten the effective foot length. Respective roll over shapes are overlayed onto each foot, which were obtained using quasi-static testing (Hansen et al. 2006).

The prosthetic feet were then modified two times: 1) by creating a wedge cut at 70% of the foot length (MEDIUM arc length foot), and then 2) by creating a second wedge cut at 60% of the total foot length (SHORT arc length foot), effectively shortening the roll over shape arc lengths (Figure 5.1). In previous research, quasi-static testing of feet with similar modifications made to them had effective foot length ratios (EFLR) of 0.62, 0.74, and 0.82 for the Short, Medium, and Long foot, respectively (Hansen et al. 2006).

These EFLRs spanned a range of those for commercially available feet (Hansen, Sam et al. 2004). Subjects were allowed to walk in the laboratory with the new foot condition until they felt comfortable (approximately 5 to 10 minutes). They were again asked to walk at self-selected slow, normal, and fast speeds for each foot type.

No changes in alignment were performed between these foot types. Subjects doffed their prostheses between conditions, and modifications were performed in another laboratory, with no alterations made besides making a wedge cut in the prosthetic foot. A total of 9 different walking conditions (3 roll over shape arc lengths x 3 speeds) were performed. Walking trials were repeated until 3-5 clean force platform hits were obtained for each leg. Subjects were allowed to rest in between trials as needed.

After all data were collected, the prostheses were restored to their original condition (i.e. subject's original foot and pylon were reattached to the socket in its original configuration), and returned to the user.

5.2.2 Data analyses

Data were processed using Motion Analysis' EVa and Orthotrak software. Missing data points in marker position data were interpolated using a cubic spline technique. Raw marker position data were filtered using a fourth-order bidirectional Butterworth infinite-impulse response digital filter with an effective cutoff frequency of 6.0Hz. Data were further processed using custom programs in Microsoft Excel (Microsoft Corporation, Redmond, WA) and Matlab (The Mathworks, Inc, Natick, MA).

Parameters studied include walking speed, step length, cadence, and maximum external ankle dorsiflexion moments during the stance phase of gait. Vertical ground reaction force 1st and 2nd peak values and timing, as well as the difference between 1st and 2nd peak values were also analyzed. Ankle-foot roll over shapes were created, and the effective foot length ratio (EFLR) was reported. The EFLR is a measure of the total foot length used during a walking step. It is measured as the effective foot length divided by the total foot length and multiplied by 100. Effective foot length was measured as the distance from the heel of the foot to the anterior end of the ankle-foot roll over shape divided by the overall foot length (Hansen, Sam et al. 2004). Left and right limbs displayed fairly similar walking patterns, so only left side data were used in the analyses.

5.2.3 **Statistical analysis**

Data were checked for normality using the Shapiro-Wilk test of Normality. Mean values of each of the data sets for each subject were used. Nine total walking conditions were analyzed (3 speeds and 3 foot types). 3 x 3 two-way repeated measures ANOVA tests were used to compare data sets for the three walking speeds (slow, normal, and fast) with the three foot types (Long, Medium, and Short), and for the interaction between speed and foot type.

Mauchly's Test of Sphericity was performed on each data set to test assumptions of the ANOVA test. When the assumption of sphericity was violated, the Greenhouse-Geisser correction factor was used to determine the P value. Pairwise comparisons were made using Bonferroni adjustments for multiple comparisons when the data were found to be significant. SPSS software (SPSS Inc, Chicago, IL) was used to perform the statistical analyses. The level of statistical significance for each test was set at a value of $P<0.05$.

5.3 Results

5.3.1 Subject vital information and temporospatial data

Of the 12 subjects tested, one subject was not able to walk at the slow walking conditions without using an assistive device (i.e. a cane). Thus, data from only 11 subjects were used in the data analyses. These subjects consisted of 5 females and 6 males with mean age, mass, and height of 52.8 years, 81.0 kg, and 169.1 cm (Table 5.1).

Table 5.1: Subject's Vital Statistics

Subject	Cause of amputation	Gender	Age (yrs)	Mass (kg)	Height (cm)
1	infection	F	44	75	162
2	disease	F	61	61	157
3	diabetes mellitus	F	47	67	168
4	infection	M	55	92	171
5	trauma	M	50	75	176
6	trauma	F	66	109	170
7	diabetes mellitus	M	30	51	173
8	congenital	M	68	128	182
9	infection	M	64	93	178
10	trauma	F	30	87	157
11	infection	M	66	55	168
Mean (SD)		-	**52.8 (13.9)**	**81.0 (23.6)**	**169.1 (8.0)**

As expected, temporospatial properties of speed, step length, and cadence (Table 5.2) were significantly different ($P<0.001$) for the different walking speed conditions (slow, normal, and fast). No significant differences in these measurements were observed for

Walking Condition	Mean Speed (m/s)	Mean Step Length (m)	Mean cadence (steps/min)	Double support time (% gait cycle)
LONG foot, Slow	0.56 (0.24)	0.44 (0.14)	73.7 (13.5)	20.1 (7.8)
MEDIUM foot, Slow	0.53 (0.25)	0.42 (0.15)	71.9 (13.1)	20.6 (7.7)
SHORT foot, Slow	0.53 (0.25)	0.42 (0.15)	72.9 (13.0)	20.9 (7.8)
LONG foot, Normal	1.00 (0.18)	0.59 (0.11)	101.6 (5.5)	14.0 (3.9)
MEDIUM foot, Normal	0.97 (0.19)	0.58 (0.10)	100.2 (5.9)	14.6 (4.0)
SHORT foot, Normal	0.99 (0.16)	0.58 (0.09)	102.1 (7.9)	14.7 (3.3)
LONG foot, Fast	1.37 (0.34)	0.68 (0.14)	118.9 (14.6)	11.9 (4.2)
MEDIUM foot, Fast	1.39 (0.34)	0.68 (0.14)	121.6 (17.9)	12.2 (3.9)
SHORT foot, Fast	1.38 (0.31)	0.67 (0.13)	123.2 (17.9)	12.4 (3.6)

Table 5.2: Temporospatial Data

prosthetic foot arc length (Short, Medium, and Long) when controlling for speed. An increase in walking speed of 0.86 m/s from the slowest to the fastest mean speeds was observed. No significant difference in speed was observed for the different foot types (P=0.735), or the interaction between foot type and speed (P=0.539). Step length and cadence increased significantly for increased walking speed (P<0.001), but not for changing foot type (P=0.136 and P=0.353, respectively) or the interaction between foot type and speed (P=0.539 and P=0.232, respectively). There was a trend observed for increasing step length as foot type went from Short to Long.

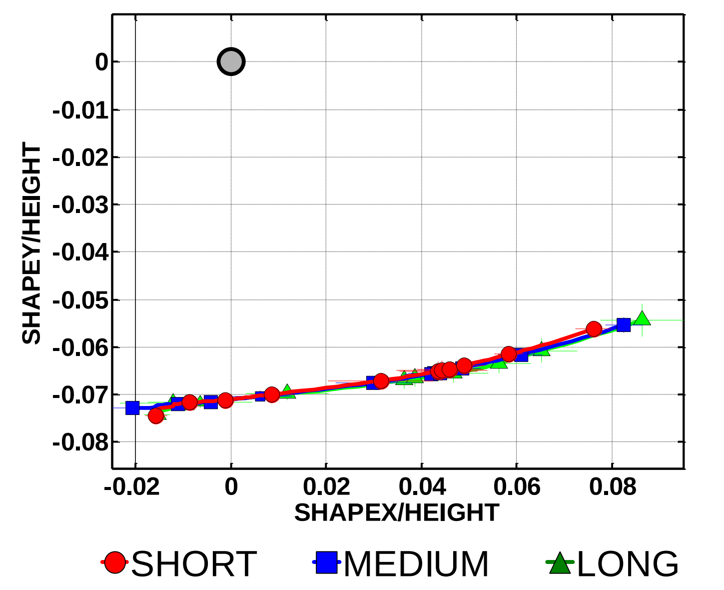

Figure 5.2: Representative ankle-foot roll over shape arc lengths for subject walking at self-selected normal speed with short, medium and long foot types.

5.3.2 Ankle-foot roll over shape arc length

Mean ankle-foot roll over shapes were calculated for each of the subjects for the nine different walking conditions. Representative roll over shapes of one subject walking at freely-selected walking speed with the three different arc length feet are plotted in Figure 5.2. There were few differences between the foot shapes for the different foot types. Ankle-foot roll over shapes also looked similar for the different walking speeds, with the exception of the

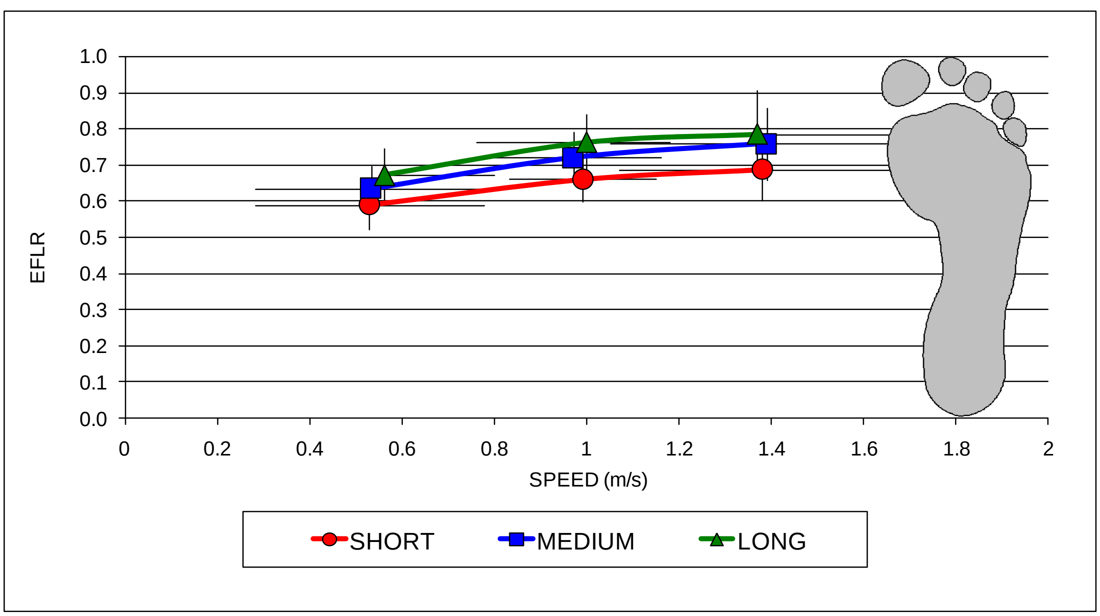

Figure 5.3: Mean Effective Foot Length Ratio for 11 subjects with bilateral trans-tibial amputation walking at three different foot types for slow, normal, and fast walking speeds.

forefoot arc length. EFLR measurements are plotted in Figure 5.3. Significant differences were found for EFLRs with prosthetic foot type (P<0.001) and speed (P<0.001), but not for the interaction between foot type and speed (P=0.083). Pairwise comparisons showed the EFLR was significantly different between all foot types at all speeds (P<0.05). EFLRs were greater for the longer foot arc length conditions and for faster walking speeds. These differences indicate that changes made to the Shape&Roll foot did effectively shorten the arc length of the foot.

5.3.3 Kinetic results

Mean stance peak ankle moment curves were examined for each subject. Representative plots of ankle flexion moment (Figure 5.4) indicate how peak ankle dorsiflexion moments increase with both walking speed and roll over shape arc length of the foot. Peak ankle dorsiflexion moment (Figure 5.5) significantly increased from Short to Long arc length (P<0.001), and as speed increased from slow to fast (P<0.001). The interaction between foot type and speed was also significant (P=0.03). Pairwise

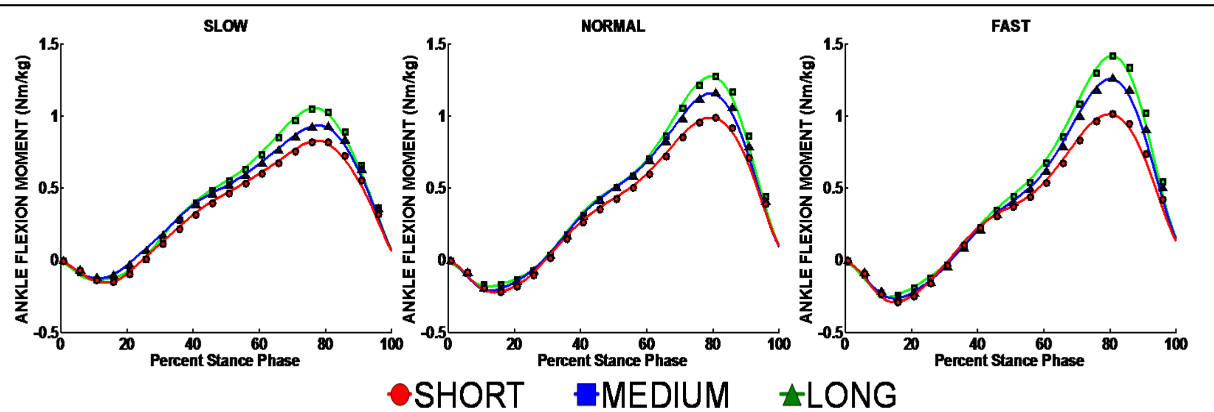

Figure 5.4: Mean dorsiflexion ankle moment curves for 11 subjects walking with three different foot types at slow, normal, and fast walking speeds

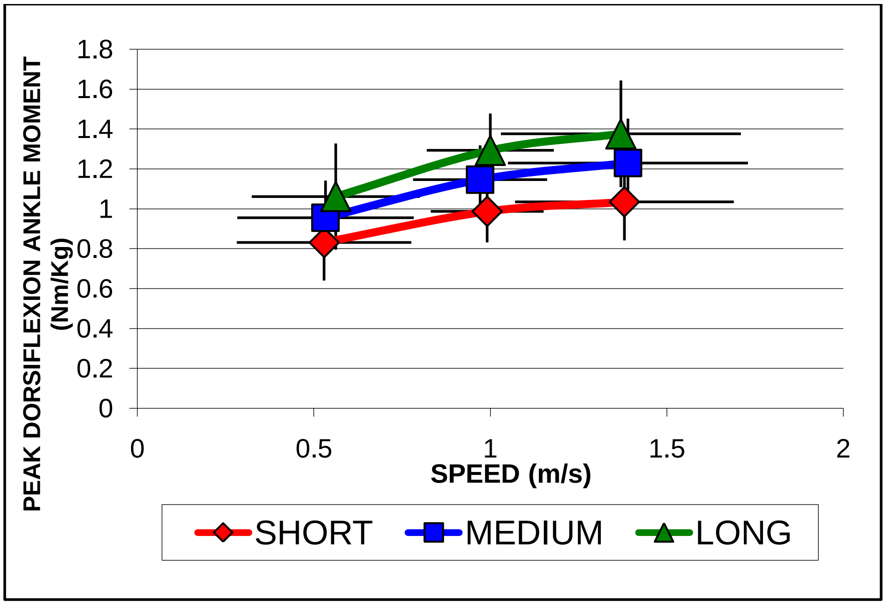

Figure 5.5: Mean values of peak dorsiflexion ankle moment for 11 subjects walking with three different foot types at slow, normal, and fast walking speeds

comparisons indicated that peak ankle moment was significantly different (P<0.05) between all speeds and foot types.

Vertical ground reaction force curves (Figure 5.6) were also analyzed. No significant differences were found for left side vertical ground reaction force (vGRF) 1st peak between all three different foot types at each speed (P=0.179). 1st peak vGRF significantly increased for increasing speed (P=0.037) and for the interaction between foot type and speed (P<0.001). Pairwise comparisons found significant differences between pairs of feet, namely, the medium and short arc length feet during the fast walking speed condition (P=0.015) and between the long and short arc length feet during the normal (P=0.050) and slow (P=0.026) walking conditions.

Left side vGRF 2nd peak was significantly higher for longer arc lengths (P<0.001), but was not significantly different between speed conditions (P=0.212). The values of the vGRF 2nd peaks were significantly different for the interaction between foot type and speed (P<0.001). Pairwise comparisons indicated that there were significant differences between all foot types for fast walking speeds (P=0.025, 0.001, and 0.009 between Long and Medium, Long and Short, and Medium and Short foot types, respectively) and between the Long and Medium (P=0.023) and Long and Short (P=0.006) foot types for normal walking speeds. No significant differences were found at the slow walking speeds between foot types. The timing of the vGRF 2nd peak was found to be significantly different for both foot type (P=0.002) and speed (P<0.001), though not for the interaction between foot type and speed (P=0.522). Timing of the vGRF 2nd peak occurred later in the stance phase of gait as roll over arc length of the foot increased and as walking speed increased.

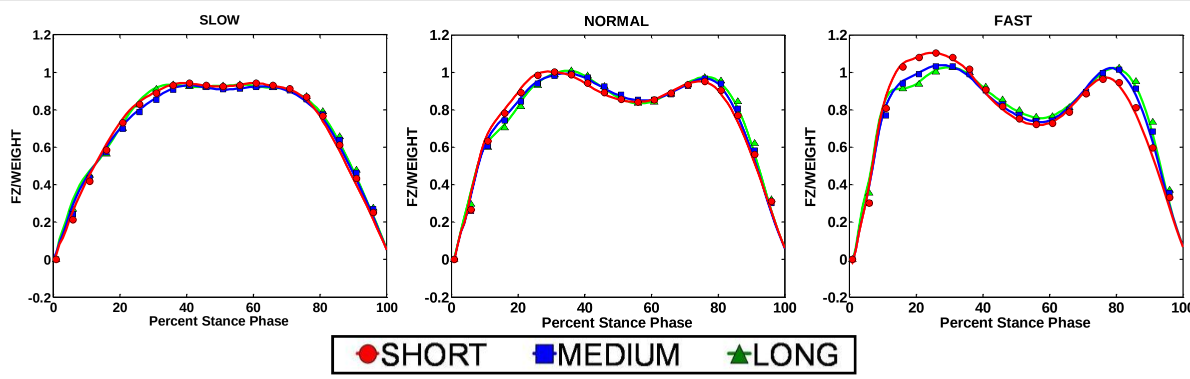

Figure 5.6: Plots of mean vertical ground reaction forces for 11 subjects walking at slow (left), normal (middle), and fast (right) speeds for short, normal, and long foot types.

Differences between the 1st and 2nd peaks of the vGRF were also calculated (1st peak minus 2nd peak) to obtain a measure of the "drop-off" occurring from the trailing limb onto the leading limb (Figure 5.7). At the slow walking speeds, mean values of the 11 subjects were highest for the long arc length foot, while the mean value was highest for the short arc length foot at the fast walking speeds. The largest difference between the 1st and 2nd vGRF peaks of 0.21 was measured for the short arc length foot at the fast walking condition. For normal walking, the difference in peak values of approximately 0.1 Nm/Kg for all foot types was observed. 1st and 2nd peak differences were not found to be significant for foot type (P=0.100) or speed (P=0.215), but were significant for the interaction between foot type and speed (P<0.001). Pairwise comparisons found that significant differences occurred between the long and short arc length foot (P=0.004) and medium and short arc length foot (P=0.002) during fast walking, and between the long and short arc length foot (P=0.046) during slow walking.

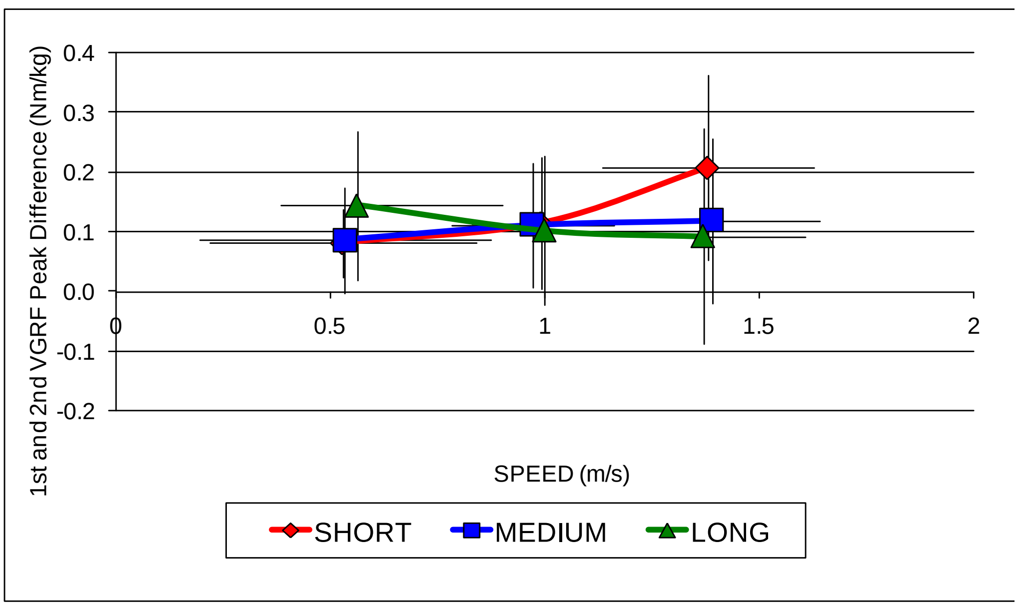

Figure 5.7: Mean difference in 1st and 2nd VGRF peaks for the three different arc length feet walking at slow, normal, and fast speeds.

An analysis of fore-aft GRF (Figure 5.8), found that left side fore-aft GRF 2nd

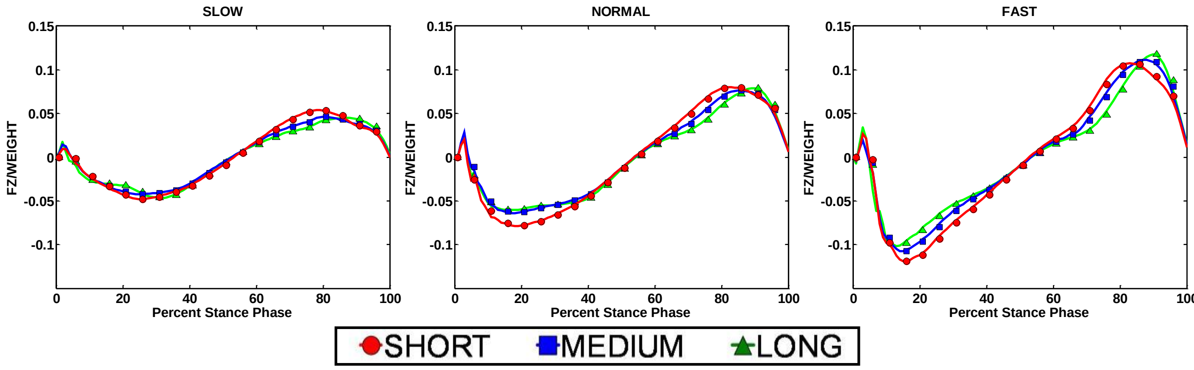

Figure 5.8: Plots of mean fore-aft ground reaction forces for 11 subjects walking at slow (left), normal (middle), and fast (right) speeds for short, normal, and long foot types.

peaks were not significantly different between foot types (P=0.696), but were found to be significantly higher for increasing speed (P=0.002). The interaction between foot type and speed was also significant (P=0.009), though pairwise comparisons found no significant differences between the three foot types or the three speeds. The timing of the fore-aft GRF 2nd peaks as a percentage of stance phase was not found to be significant for foot type (P=0.165) or the interaction between foot type and speed (P=0.077). Fore-aft GRF 2nd peak timing occurred significantly later in the stance phase of gait for faster walking speeds (P=0.016).

5.4 Discussion

The range of values of the Effective Foot Length Ratios measured for subjects with bilateral trans-tibial amputation in this study (0.6 – 0.8) were similar to those found in quasi-static prosthetic foot characterization of these feet (0.6 – 0.83) (Hansen et al, 2004), but were lower than that measured during dynamic walking of persons with unilateral trans-tibial amputation walking with similar Shape&Roll feet (Hansen et al, 2006). Maximum EFLR of the unilateral subjects was 0.9. Significant differences in

EFLR and peak ankle dorsiflexion moment show that gait changes do occur with different effective roll over shapes, but bilateral prostheses users did not modify their gait characteristics in the same way as that observed in persons with unilateral trans-tibial amputation (Hansen et al, 2006). Namely, no significant differences in 1st peak vertical ground reaction force were observed between all three foot types (though pairs of feet were found to be significantly different from each other for certain walking speed conditions).

Though the differences in 1st and 2nd peaks of the vGRF were not found to be significant for foot type or speed, the interaction between foot type and speed was significant. For the short arc length foot, the difference between the peaks increased with speed. The vGRF 2nd peak was found to be significantly lower for shorter arc length feet. The results suggest that differences in vGRF patterns occur for short roll over shape arc lengths at fast walking speeds. Under these conditions, the COP progresses to the end of the forefoot and a “drop-off” onto the contralateral limb occurs more abruptly as body mass is transferred from the trailing to the leading leg, causing increased vertical loading during early stance. When the walking speed is lower or the roll over shape arc length is longer (e.g. walking with the Medium and Long foot types), the COP does not progress past the end of the “useable” forefoot, so no “drop-off” effect is observed. The smaller EFLRs for the medium and long arc length feet compared to that measured for the same feet on subjects with unilateral trans-tibial amputation (Hansen et al. 2006) also suggest that the persons in our study did not utilize the full roll over shape arc length potential of the feet. The slower walking speeds of the bilateral subjects (0.5 - 1.4 m/s) compared to the unilateral subjects (0.8 - 1.7 m/s) may be a cause of the decreased roll over shape arc lengths observed in the subjects in this study. Step length values were not reported for the unilateral group, so it is difficult to determine if shorter step lengths were observed in the bilateral subjects compared to the unilateral subjects, though step lengths during normal walking were approximately 0.10m shorter than able-bodied freely selected step lengths.

It was expected that changing the roll over shape arc length of the prosthetic foot would have an effect on temporospatial gait parameters. Though there was a trend for increased step length for longer arc length feet, there were no significant differences in speed. This seems to be due to the fact that there was no pattern relating cadence with foot type and the three walking speeds. Of interest, cadence increased at fast walking speeds as the roll over shape arc length decreased. Though this could be due to subjects spending less time in stance because the center of pressure progresses to the end of the useable forefoot sooner, it also allows them to take these steps faster.

During testing, subjects were asked which foot type they preferred. Most subjects seemed to prefer the foot type that had a roll over shape arc length most similar to that of their original foot (previously measured from quasi-static testing by Hansen et al (2004)). Generally, those who were more active liked the long and medium arc length feet, while those who walked slower preferred the medium or short arc length feet. This may also be a reason why EFLRs were shorter, as many of the subjects were originally fit with prosthetic feet having shorter roll over shape arc lengths. More differences might be observed if subjects were able to walk faster, as EFLR would most likely increase for the longer arc length feet. It is possible though, that without having a sound limb to “catch” themselves as they transitioned from prosthetic stance phase during walking, subjects chose slower walking speeds so that ground reaction force peaks were reduced to prevent discomfort or injury. Further investigations of bilateral trans-tibial walking and the effects of different prosthetic components, training, and alignment are needed to better understand where improvements to gait of persons with amputation can be made.

Chapter 6: Comparison of step length of able-bodied persons and persons with bilateral trans-tibial amputation

6.1 Introduction

It is believed that the ankle foot complex of the trailing limb plays an important role in gait characteristics, specifically, step length. Step length is significantly shorter in persons with bilateral lower limb amputations (0.57 m) compared to step lengths of able-bodied persons (0.69 m) when comparing freely selected walking speeds (Su et al. 2007). At comparable speeds around 0.9 m/s (freely-selected walking of bilateral amputee subjects versus slow walking of able-bodied subjects), step lengths were not significantly different, but the differences in kinematics between the two groups suggest that gait strategies are different to achieve this step length.

By changing the forefoot arc length of the effective foot rocker on persons with bilateral amputation, it is possible to investigate its effect on gait characteristics, particularly step length. It is believed that the roll over shape arc length plays a role in gait of persons with amputation, since it has been observed that prosthetic feet have shorter effective rocker arc lengths than the ankle-foot systems of able-bodied persons (Hansen, Sam et al. 2004). In a study of persons with unilateral trans-tibial amputation, significantly slower walking speeds and an observed increase in step length asymmetry were measured when the arc length of the prosthetic foot was shortened (Hansen et al.

2006). A similar study in persons with bilateral trans-tibial amputation will eliminate possible effects of compensation by the sound limb.

Since a range of different effective foot lengths have been measured (using quasi-static testing) for different prosthetic feet (Hansen, Sam et al. 2004), a study to determine if there are differences in gait characteristics due to different roll over shape arc lengths may be helpful in deciding which prosthetic feet would be most beneficial for persons with bilateral trans-tibial amputation. Also, very few studies have examined gait characteristics of persons with bilateral trans-tibial amputation (Su et al. 2007; 2008; Tsai et al. 2003). This study will also aim to improve upon the knowledge of bilateral trans-tibial amputee gait.

6.2 Methods

6.2.1 Data acquisition

Data were previously acquired from 11 subjects with bilateral trans-tibial amputation (BTTA) who walked with endoskeletal prostheses. They signed consent forms that were approved by Northwestern University's Institutional Review Board. Data collection and analyses for the study were conducted in the VA Chicago Motion Analysis Research Laboratory (VACMARL). An eight-camera Eagle Digital Real-Time motion measurement system (Motion Analysis Corporation, Santa Rosa, CA) was used to acquire marker movements at 120 Hz and calculate kinematic data. Ground reaction forces were acquired using six AMTI (Advanced Mechanical Technology, Inc., Watertown, MA) force platforms simultaneously recorded with the motion analysis cameras at 960 Hz. A modified Helen Hayes marker set (Kadaba et al. 1990) was used to define a biomechanical model of each person. Markers for the ankle, heel, and toe were placed on a specialized plate between the pylon and the foot, at the level of the ankle. This was used to allow for possible comparison between studies and to prevent variability in measurement due to changes in marker placement when changing the foot type. A static standing trial was performed before the walking experiment in order to estimate the location of the joint centers of rotation.

A detailed protocol is discussed in Chapter 5 and is similar to that described by Hansen et al. (2004; Hansen et al. 2006). Subjects walked at self-selected normal, fast, and slow speeds while wearing three different prosthetic feet. The first (LONG) foot was an unaltered Shape&Roll prosthetic foot. The foot is made of a copolymer plastic that is made to conform to the roll over shape (effective rocker shape that is formed while walking) of the able-bodied ankle-foot system during walking (Sam et al. 2004). Changes can be made in the roll over arc length without making changes to other prosthetic parameters. The second (MEDIUM) foot tested was the same Shape&Roll prosthetic foot with a wedge cut made in it at 70% of the total foot length. The third (SHORT) foot was the same Shape&Roll prosthetic foot with a second wedge cut made in it at 60% of the total foot length. These wedge cuts effectively shortened the roll over shape arc length of the foot. Wedge cut locations were chosen to span the roll over shape arc lengths of various commercially available prosthetic feet as measured in our laboratory from quasi-static testing (Hansen, Sam et al. 2004). The feet were fit onto the subject's original sockets with rigid pylons. The subjects were blinded to the type of changes made to the feet during the study. Dynamic alignment was performed by a qualified prosthetist before subjects walked with the first foot, but no alignment changes were made between foot types. Subjects doffed their prostheses between conditions and modifications were performed in another laboratory, with no alterations made besides making the wedge cuts in the feet. A total of 9 different walking conditions (3 different roll over shape arc lengths x 3 different speeds) were performed. Walking trials were repeated until 3-5 clean force platform hits were obtained for each leg. Subjects were allowed to rest in between trials as needed.

Data were previously acquired from 10 able-bodied subjects (mean age, mass, and height was 24 ± 1 years, 69.3 ± 9.5 kg, and 174.6 ± 9.1 cm) walking at freely-selected walking speeds and were compared with the BTTA subjects.

6.3 Data analyses

Data were processed using Motion Analysis' EVa and Orthotrak software. Missing data points in marker position data were interpolated using a cubic spline technique. Raw marker position data were filtered using a fourth-order bidirectional Butterworth infinite-impulse response digital filter with an effective cutoff frequency of 6.0Hz. Data were further processed using custom programs in Microsoft Excel (Microsoft Corporation, Redmond, WA) and Matlab (The Mathworks, Inc, Natick, MA). It was assumed that subjects walked fairly symmetrically on both sides. Therefore, data for this step length analysis were calculated from only left side steps.

Comparisons were performed between the group of persons with bilateral trans-tibial amputations and able-bodied persons. Analyses were run on both groups' data for freely-selected walking speeds, and also on data where the two groups' speed and step length were matched (no statistically significant differences between the two). Comparisons were made between the BTTA group walking with each of the 3 different feet and the able-bodied subjects.

Specific gait data that were analyzed for this study were walking speed, step length, cadence, and double limb support time for all walking conditions. A Segment Contribution to Step Length (SCSL) analysis was also performed. The SCSL analysis calculates the contribution of six lower limb segments to the overall step length. The segments that were included were the trailing ankle-foot, shank, and thigh segments, the pelvic segment, and the leading thigh and shank segments (Figure 6.1). Specifics of how each segment contribution is calculated are provided in Chapter 2. Measurements of each segment were reported as a percentage contribution to overall step length.

Comparisons in SCSL were also performed on the bilateral trans-tibial amputee subjects walking with the three different arc length feet at the three walking speeds.

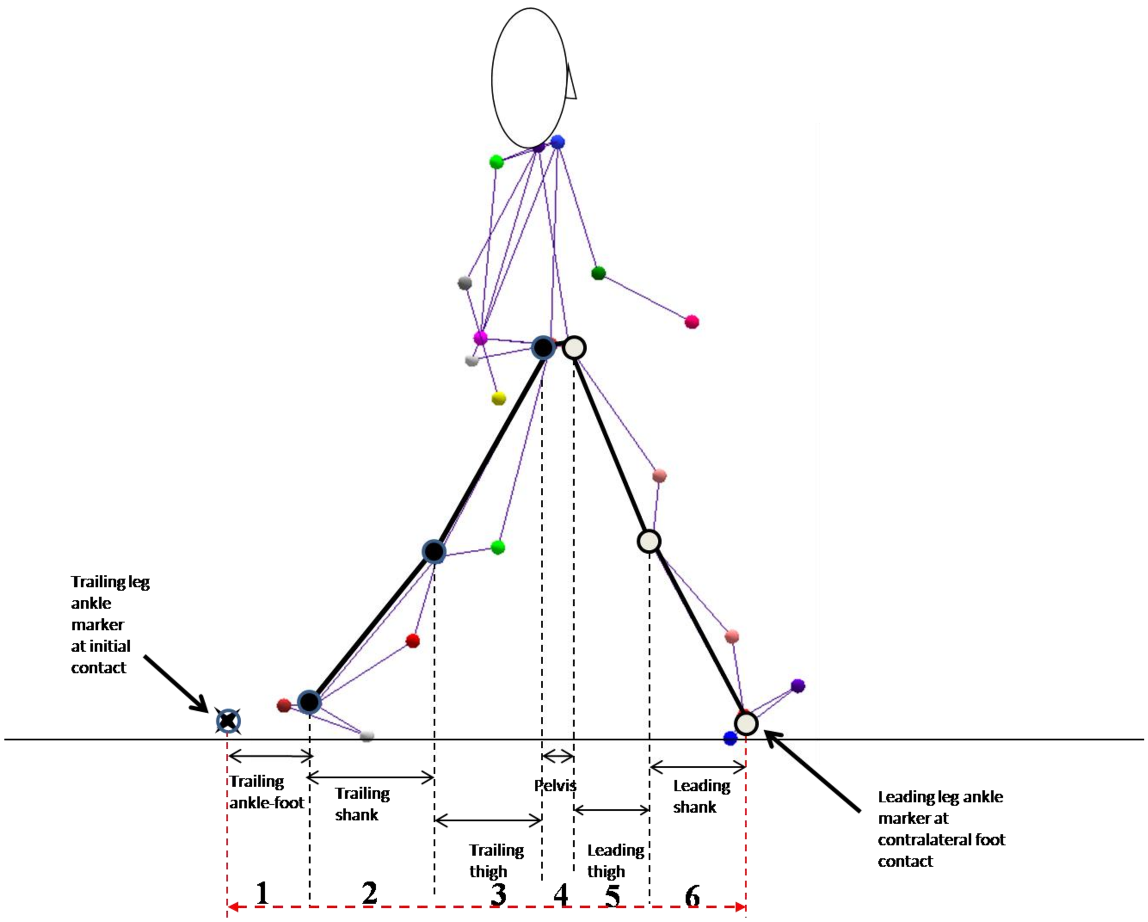

Figure 6.1: Representation of segments used for Segment Contribution to Step Length analysis. Segment 1, the trailing ankle-foot segment, is the sagittal distance traversed by the trailing ankle joint center from foot contact to opposite foot contact. Segments 2-6 (shank, thigh and pelvis segments) are the measured lengths of these segments at the time of initial contact of the leading limb. The lengths of each shank segment is measured from ankle joint center to knee joint center, while the thigh segments are measured from knee joint center to hip joint center. The pelvic segment length is measured as the distance between the two hip joint centers.

6.3.1 **Statistical analyses**

Data were checked for normality using the Shapiro-Wilk test of Normality. The level of statistical significance for each test was set at $P<0.05$. Statistical analyses were performed between the group of persons with bilateral trans-tibial amputation and group of able-bodied persons when walking at freely-selected speeds and also at speeds matched between the two groups. One way analysis of variance was used with Bonferroni correction. Mean values for each of the subjects were used in the analysis.

Statistical analyses were also performed on the data of bilateral trans-tibial amputee subjects walking with the three different arc length feet at three different walking speeds. Mean values of each of the data sets for each subject were used. Nine total walking conditions were analyzed (3 speeds and 3 foot types). 3 x 3 two-way repeated measures ANOVA tests were used to compare data sets for the three walking speeds (slow, normal, and fast) with the three foot types (Long, Medium, and Short), and for the interaction between speed and foot type. Mauchly's Test of Sphericity was performed on each data set to test assumptions of the ANOVA test. When the assumption of sphericity was violated, the Greenhouse-Geisser correction factor was used to determine the P value. Pairwise comparisons were made using Bonferroni adjustments for multiple comparisons when the data were found to be significant. The level of significance was set at a value of $P<0.05$. SPSS software (SPSS Inc, Chicago, IL) was used to perform all statistical analyses.

Statistical analyses were performed for walking speed, step length, cadence, and double support time. Percent contribution to overall step length of each lower limb segment (trailing ankle-foot, trailing shank, trailing thigh, pelvis, leading thigh, and leading shank), as well as contribution by the leading and trailing limbs, were also compared.

6.4 Results

6.4.1 Subject information and temporospatial data

Data from 6 males and 5 females with bilateral trans-tibial amputation (mean age, mass, and height of 53 years, 81.0 kg, and 169.1 cm, respectively,) and 10 able-bodied subjects (5 males and 5 females with mean age, mass, and height of 25 years, 69.0 kg, and 174.6 cm, respectively,) were used in this analysis. Vital statistics of both groups are displayed in Table 6.1. The age of the able-bodied group was significantly younger than the amputee group. No significant differences were found between the height and weight of the two groups.

Table 6.1: Subject's vital statistics

Subject	Cause of amputation	Gender	Age (yrs)	Mass (kg)	Height (cm)
BTT Amputee Group					
1	infection	F	44	75	162
2	disease	F	61	61	157
3	diabetes mellitus	M	66	55	168
4	infection	F	47	67	168
5	trauma	M	55	92	171
6	trauma	M	50	75	176
7	diabetes mellitus	F	66	109	170
8	congenital	M	30	51	174
9	infection	M	68	128	182
10	trauma	M	64	93	178
11	infection	F	30	87	157
Mean (SD)		**-**	**52.8 (13.9)**	**81.0 (23.6)**	**169.1 (8.0)**
Able Bodied Group					
1	-	M	26	86	192
2	-	F	23	60	171
3	-	F	24	58	165
4	-	M	24	71	170
5	-	F	24	58	171
6	-	F	23	65	164
7	-	M	24	72	181
8	-	F	27	81	169
9	-	M	26	70	178
10	-	M	26	73	186
Mean (SD)			**24.7 (1.4)**	**69 (10)**	**174.6 (9.1)**

Temporospatial parameters of the subjects with bilateral trans-tibial amputation were compared with that of the able-bodied group (Table 6.2). Mean freely selected walking speed, step length, and cadence of the BTTA group were similar between the different foot types, and were around 0.99 m/s, 0.59 m, and 101 steps/min, respectively. When comparing the BTTA and able-bodied groups, significant differences were found for speed, step length, and cadence for all foot types when walking at freely-selected walking speeds. The able-bodied group walked significantly faster and with longer step lengths compared with the BTTA group for all arc length feet (P≤0.001 for all feet) during freely-selected walking. Able-bodied persons also walked at significantly higher cadences than the BTTA group for long (P=0.012), medium (P=0.005), and short (P=0.035) arc length feet. Double support time was significantly shorter for able-bodied

	BTTA LONG (1)			BTTA MEDIUM (2)			BTTA SHORT (3)			ABLE-BODIED (4)	Significance (between AB and BTTA groups)	
Measure	**Fast**	**Free**	**Slow**	**Fast**	**Free**	**Slow**	**Fast**	**Free**	**Slow**	**Free**	**Free**	**Matched**
Speed (m/s)	1.37	1.00	0.56	1.39	0.97	0.53	1.38	0.99	0.53	1.34	1-4 2-4 3-4	none
Step Length (m)	0.69	0.59	0.44	0.69	0.58	0.42	0.67	0.59	0.42	0.73	1-4 2-4 3-4	none
Cadence (step/min)	118.9	101.6	73.7	121.6	100.2	71.9	123.2	102.0	72.9	110.1	1-4 2-4 3-4	3-4
Double-support time (% gait cycle)	11.8	14.0	20.1	12.2	14.6	20.6	12.4	14.7	20.9	10.7	1-4 2-4 3-4	none

Table 6.2: Temporospatial data of persons with bilateral trans-tibial amputation (BTTA) and able-bodied persons. Persons with BTTA walked at three speeds with three different feet: Long arc length foot (1), Medium arc length foot (2), and Short arc length foot (3). These groups were compared for significance between the able-bodied group walking at their freely-selected walking speed (4). Matched walking speed comparisons were run between the BTTA fast walking speeds and the able-bodied freely-selected walking speed (shaded values). Statistical significance is noted in the right hand column for both freely selected walking speed and also when walking speeds were matched. As an example: "1-4" indicates there were significant differences between groups 1 and 4.

persons compared to the BTTA group for long (P=0.02), medium (P=0.009), and short (P=0.002) arc length feet.

Similar walking speeds (approximately 1.38 m/s) to the able-bodied freely-selected speed of 1.34 m/s occurred for the amputee group when they were asked to walk fast. When the data were speed-matched, significance in temporospatial data was only found for cadence between BTTA subjects walking with the short arc length foot and the able-bodied subject group (P=0.048). Mean cadence of the BTTA group was higher than the able-bodied group and increased as the roll over shape arc length of the foot decreased. At these speeds, step length of the bilateral amputee subjects was around 0.68 m compared to 0.73 m of the able-bodied group, though these differences were not significant (P=0.298, P=0.290, and P=0.172, for long, medium, and short arc length feet, respectively).

6.4.2 Segment Contribution to Step Length (SCSL) Analysis: Comparisons between persons with bilateral trans-tibial amputation and able-bodied persons

Though the values of the mean percent contribution of each of the lower limb segments were different between the walking speeds of the BTTA group, significant differences in percent contribution by each segment compared to that for able-bodied walking were similar for both freely selected walking speed comparisons and speed matched walking (Table 6.3). Total contribution of each segment to step length for the BTTA group and the able-bodied group are plotted in Figure 6.2. In some instances, the pelvic

	BTTA LONG (1)			BTTA MEDIUM (2)			BTTA SHORT (3)			ABLE-BODIED (4)	Significance (between AB and BTTA groups)	
Measures	**Fast**	**Free**	**Slow**	**Fast**	**Free**	**Slow**	**Fast**	**Free**	**Slow**	**Free**	**Free**	**Matched**
% Trail Ankle-Foot Contribution	16.0	15.7	15.4	16.2	15.8	15.4	16.1	16.2	15.1	14.8	3-4	2-4,3-4
% Trail Shank Contribution	21.8	22.8	25.4	21.7	22.5	25.5	21.2	22.2	25.0	21.4	none	none
% Trail Thigh Contribution	14.5	13.3	10.7	13.9	13.0	8.8	12.8	11.7	9.6	20.2	1-4, 2-4, 3-4	1-4, 2-4, 3-4
% Pelvis Contribution	0.6	-0.6	-3.5	0.9	-0.7	-3.5	0.9	-0.1	-3.1	0.3	None	none
% Lead Thigh Contribution	26.2	28.7	35.2	26.4	28.6	38.2	27.3	28.7	37.5	18.9	1-4, 2-4, 3-4	1-4, 2-4, 3-4
% Lead Shank Contribution	20.9	20.2	16.9	20.9	20.8	15.6	21.6	21.3	16.0	24.8	1-4, 2-4, 3-4	1-4, 2-4, 3-4

Table 6.3: Percent contribution of the lower limb segments to step length of persons with bilateral trans-tibial amputation (BTTA) and able-bodied persons. Persons with BTTA walked at three speeds with three different feet: Long arc length foot (1), Medium arc length foot (2), and Short arc length foot (3). These groups were compared for significance between the able-bodied group walking at their freely-selected walking speed (4). Comparisons were conducted at similar speeds between groups, using the BTTA fast walking speed and the able-bodied freely-selected walking speed (shaded values). Statistical significance is noted in the right hand column for both freely selected walking speed and also when walking speeds were matched. As an example: “1-4” indicates there were significant differences between groups 1 and 4.

contribution to step length was negative. In those cases, the sum of contributions by the other five segments was actually more than 100%.

Mean contribution by the trailing ankle-foot segment (Segment 1) of the 11 BTTA subjects was slightly higher compared to the able-bodied group (15% of overall step length for the able-bodied ambulators, and 16% for the BTTA group at freely-selected and fast walking speeds). The percent contribution of the trailing ankle-foot segment to overall step length was found only to be significant between the able-bodied group and

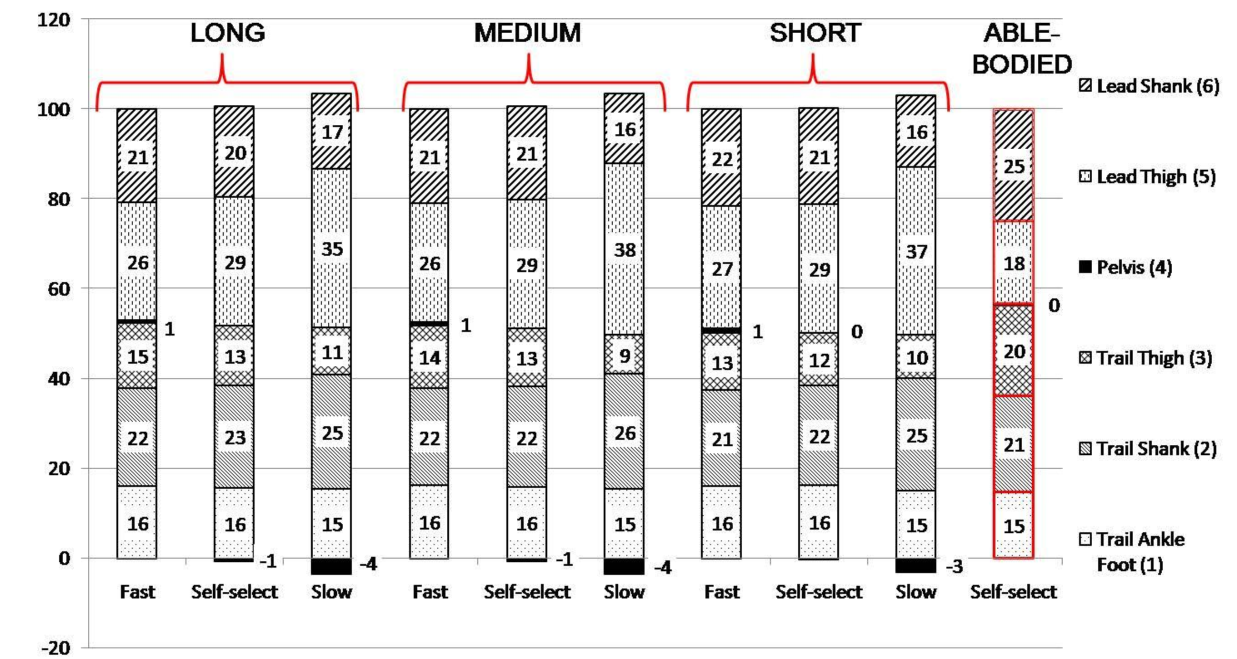

Figure 6.2: Percent contribution to overall step length of the six lower limb segments used in the segment contribution to step length analysis (see Figure 1). For the self-selected walking with the long and medium arc length feet, and for the slow walking with all arc length feet, pelvic contribution was negative.

the BTTA group walking with the short arc length foot (P=0.044) for freely-selected walking, and with the medium and short arc length foot (P=0.044 and P=0.033, respectively,) for speed matched walking. For the long arc length foot, no significant differences in contribution by this segment were observed (P=0.250 and P=0.079 for freely-selected, and speed matched walking, respectively). For the medium arc length foot, P=0.135 for freely-selected walking.

Contribution by the trailing shank segment (Segment 2) ranged between 21% and 26% of overall step length. Though it was higher in most cases in the BTTA group, there were no significant differences between the BTTA and able-bodied group for any foot type either for freely-selected or speed matched walking.

Contribution by the trailing thigh (Segment 3) for the able-bodied group was around 20% of overall step length, and between 9-13% for the BTTA group. The percent contribution of the trailing thigh segment to overall step length was found to be significantly higher for the able-bodied group compared to the BTTA group for all foot types. Significance was found for both freely-selected and speed matched walking for the long arc length foot (P=0.011 and P=0.012, respectively), medium arc length foot (P=0.009 and P=0.01, respectively), and short arc length foot (P=0.003 for both speeds).

There was very little contribution to overall step length by the pelvis (Segment 4), with mean value of 0.3% for the able-bodied group, and values ranging between -3.5% and 0.9% for the BTTA group. There were no significant differences between the BTTA and able-bodied group for any foot type either for freely-selected walking (P=0.210, P=0.201, and P=0.279 for Long, Medium, and Short foot type, respectively,) or speed matched walking (P=0.437, P=0.509, and P=0.535 for Long, Medium, and Short foot type, respectively).

Contribution by the leading thigh segment (Segment 5) for the able-bodied group was around 19% of the overall step length compared to 26-38% of the overall step length for the BTTA group. The percent contribution of the leading thigh segment to overall step length was significantly higher for the BTTA group compared to the able-bodied group for all foot types. Significance was found for both freely-selected and speed matched walking for the long arc length foot (P=0.002 and P=0.013, respectively), medium arc length foot (P=0.003 and P=0.009, respectively), and short arc length foot (P=0.002 and P=0.004, respectively).

Contribution by the leading shank segment (Segment 6) for the able-bodied group was around 25% of the overall step length compared to 17-21% of the overall step length for the BTTA group. Similar to the leading thigh segment, the percent contribution of the leading shank segment to overall step length was significantly higher for the able-bodied group compared to the BTTA group for all foot types. Significance was found for both freely-selected and speed matched walking for the long arc length foot (P=0.001 and P<0.001, respectively), medium arc length foot (P=0.003, and P<0.001, respectively), and short arc length foot (P=0.002 for both speed comparisons).

An analysis was also run to compare the trailing and leading limb contributions to step length (Figure 6.3). The trailing limb consisted of the contributions by the trailing ankle-foot, shank, and thigh segments. The leading limb was composed of the leading thigh and shank segment contributions. For able-bodied walking, contribution of the trailing limb to step length was 56% compared to 43% of the leading limb. For the BTTA group, percent contribution was near 50% for both the trailing and leading limbs. The percent contribution of the trailing limb to overall step length was significantly lower for the BTTA group compared to the able-bodied group for all foot types. Significance was found for both freely-selected and speed matched walking for the long arc length foot (P=0.01 and P=0.042, respectively), medium arc length foot (P=0.016 and P=0.018, respectively), and short arc length foot (P=0.004 and P=0.002, respectively).

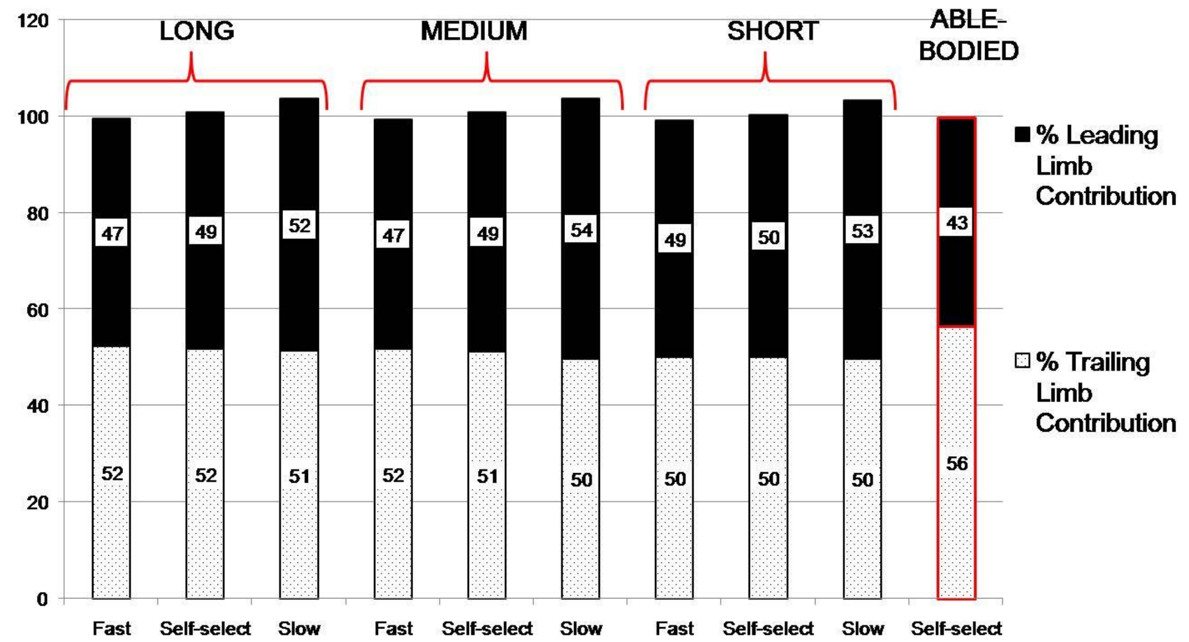

Figure 6.3: Mean percent contribution to overall step length of the trailing and leading limbs. The trailing limb contribution consists of contributions by the trailing ankle-foot, shank, and thigh segments. The leading limb contribution consists of contributions by the leading thigh and shank segments.

No significant differences in contributions to step length by the leading limb were found between the BTTA group and able-bodied group for either the freely-selected or speed matched walking conditions for the long arc length foot (P=0.057 and P=0.144, respectively), or for the speed matched condition for the medium arc length foot (P=0.101). Percent contribution by the leading limb was significantly higher for the

BTTA group compared to the able-bodied group when walking at freely-selected speeds with the medium and short arc length feet (P=0.040 and P=0.017, respectively,) and when walking at matched speeds with the short arc length foot (P=0.027).

6.4.3 Segment Contribution to Step Length (SCSL) Analysis: Comparison between different foot types and walking speeds for persons with bilateral trans-tibial amputation

Two-way repeated measures ANOVA were performed on the SCSL data to compare the three foot types and the three walking speeds of the persons with bilateral trans-tibial amputation. The differences in mean percent contribution by the trailing ankle-foot were small for the different foot types and speeds, ranging from 15% to 16%. No significance differences were found for the percent contribution to step length by the trailing ankle-foot segment for the different arc length foot types (P=0.862) or the interaction between foot type and speed (P=0.566). Significant increases in percent contribution to step length by the trailing ankle-foot segment were found for increased walking speed (P<0.001).

Though the percent contribution to step length mean value range (21% - 26%) was higher for the trailing shank segment compared to the trailing ankle-foot segment, no significant differences were found for the different arc length foot types (P=0.521), speed (P=0.081), or interaction between foot type and speed (P=0.931). There were also no significant differences for the percent contribution to step length by the trailing thigh for the different arc length foot types (P=0.100), speed (P=0.145), or the interaction between foot type and speed (P=0.479).

Significant differences were observed for percent contribution to step length by the pelvic segment (P=0.013) for walking speed. A negative contribution by the pelvis of around 4% was actually observed for slow walking speeds compared to a 1% (positive) contribution at fast speeds. No significant differences were found in the percent contribution to step length by the pelvic segment for the different arc length foot types (P=0.346) or the interaction between foot type and speed (P=0.801).

Mean values for percent contribution by the leading thigh segment ranged from 26% to 38% of overall step length for the different walking conditions of the BTTA subjects. No significant differences were found in the percent contribution by the leading thigh segment for the different arc length foot types (P=0.176) or the interaction between foot type and speed (P=0.347). Significant decreases in percent contribution by the leading thigh segment were found for increased walking speeds (P=0.001).

No significant differences were found for the percent contribution to step length by the leading shank segment for the different arc length foot types (P=0.684), speed (P=0.135), or the interaction between foot type and speed (P=0.512). It was observed that mean values during slow walking were lower (around 16%) than during freely-selected or fast walking (both around 21%).

When comparing the percent contribution by the leading and trailing limbs, significant differences were measured for the leading limb for the different arc length foot types (P=0.031) and speed (P=0.015), but not for the interaction between foot type and speed (P=0.292). Higher percent contributions to step length by the leading limb occurred for the shorter foot types and slower speeds. For trailing limb contribution, significant increases were observed for the longer arc length feet (P=0.003). No significant differences were found for speed (P=0.656) or the interaction between foot type and speed (P=0.155).

6.5 Discussion

Temporospatial data (i.e. speed, step length, cadence, and double support time) of persons with bilateral trans-tibial amputation were similar to those reported by Su et al. (2007). When comparing the BTTA data with the able-bodied data at freely-selected walking speeds, BTTA subjects' temporospatial parameters were significantly different from the able-bodied group, with slower walking speeds, step lengths, and cadences, and longer double support periods of gait. When the data were compared at similar speeds (comparing BTTA group's fast walking speeds with able-bodied freely-selected walking), step length and double support time were not significantly different. Cadence was significantly different when compared to able-bodied persons only when subjects walked with the short arc length foot at fast speeds.

Even though the subjects with bilateral trans-tibial amputation had similar step lengths (around 0.7m) as the able-bodied group when data were speed matched, significant differences were found for some of the lower-limb segments' contributions to step length, namely from the trailing and leading thigh segments and the leading shank segments. The trailing ankle-foot segment contribution was also significantly different from able-bodied walking when BTTA subjects walked with the short arc length feet at fast and freely-selected speeds, and with the medium arc length feet at fast speeds. This may suggest that the long arc length foot acts similarly to the intact ankle and foot in terms of roll over arc length, but that gait alterations occur for the shorter arc length feet to compensate for the difference in foot characteristics.

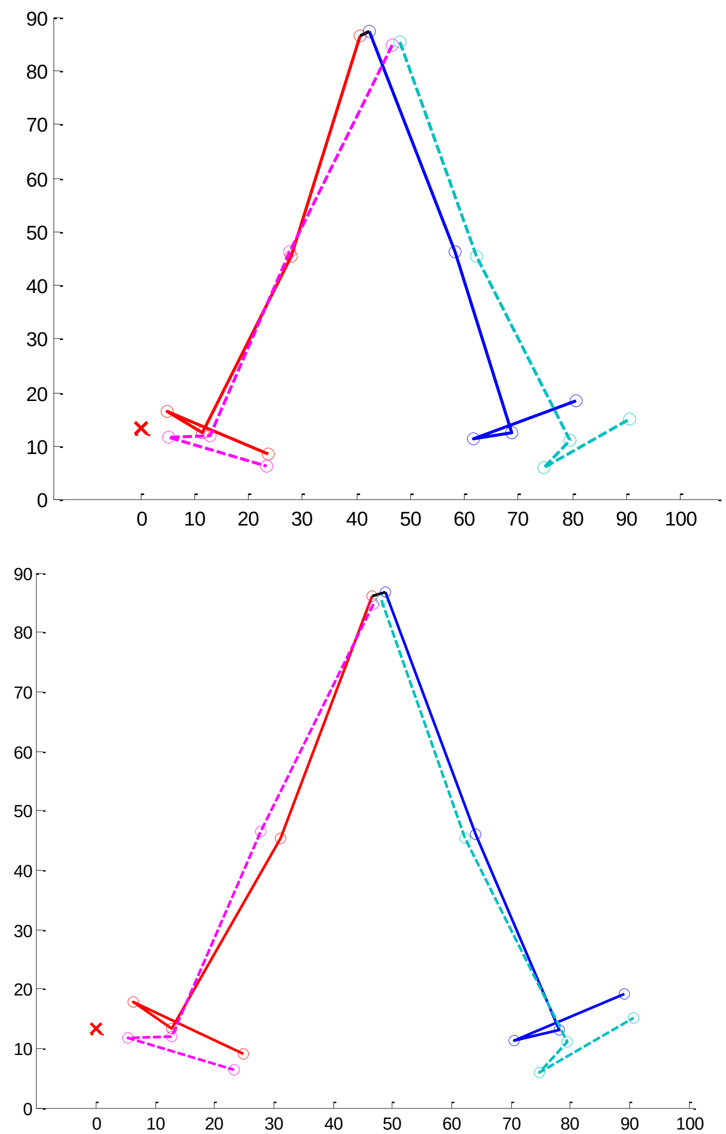

Figure 6.4: Lower limb stick figures comparing a representative able-bodied subject (dotted lines) with a representative BTTA subject (solid lines) for freely-selected walking (top) and speed matched walking (bottom). Although the step lengths are similar for the speed matched walking, each subject utilizes a somewhat different gait strategy by the lower limb segments to achieve the step length.

Data from this study suggest that, when walking at similar speeds, BTTA subjects are able to achieve comparable step lengths as able-bodied persons by using different gait strategies (Figure 6.4). Namely, higher contributions to overall step length come from the leading thigh and less from the leading shank and trailing thigh segments, compared to that of able-bodied persons. For both the trailing and leading limbs, the knees are more flexed than that of the able-bodied subjects at the time of initial contact, which play a role in these different segment contributions.

The reason for the increased knee flexion may be due to a number of different factors, including lack of ankle plantarflexor musculature to control the movement of center of progression under the foot without flexing the knee, or effects of the socket or prosthetic fit that change knee kinematics. Though Su et al. (2007) found knee flexion ROM of the BTTA subjects to be less than that of the able-bodied subjects, during speed-matched walking at the time of initial contact (when step length is measured) the BTTA subjects in our study displayed more knee flexion than able-bodied persons. Increased knee flexion on the trailing limb may increase the person's sense of stability (the limb remains closer to the body) while knee flexion in the leading limb may help the person keep the foot closer to the ground and reduce the time of swing, which may increase one's feeling of control or stability. Some differences in knee flexion could also be due to marker placement on the socket of the BTTA subjects instead of on the femoral condyles as placed on the able-bodied subjects.

As walking speed increased for the BTTA subjects, significant differences in step length were observed. Changes in percent contribution to step length by the lower limb segments were also observed; namely, significant increases in the trailing ankle-foot and pelvic segment and significant decreases in the leading thigh segment were measured for increased walking speed. The segment contribution changes were somewhat different from that of able-bodied persons when walking with larger step lengths, whose percent contribution of the trailing ankle-foot, trailing shank, pelvis, and leading thigh increased while the trailing thigh and leading shank segment contribution decreased for increases in step length (Chapter 3).

Changing the roll over shape arc length of the BTTA group did not make significant differences to the segment contributions to step length. No significant differences were observed for any of the segments for the different foot types, or the interaction between foot type and speed. When these contributions were summed in order to compare the leading and trailing limb contributions, significant increases in contribution to step length by the trailing limb were accompanied by simultaneous significant decreases in contribution by the leading limb for the longer arc length feet. This implies that walking with the different foot types caused small changes in segment contributions, which had a significant influence on the gait characteristics of the BTTA subjects. Further research needs to be conducted to determine if these differences are unfavorable for gait or cause long term deleterious effects to joints of the body. Percent contributions by the leading and trailing limbs were most similar to able-bodied data when walking with the long arc length feet, with the trailing limb contributing more to step length than the leading limb. These feet are those that would typically be prescribed for these subjects if there were fit with Shape&Roll feet, and were designed to allow similar roll over shape arc lengths to that of able-bodied persons.

It is also interesting to note that pelvic contribution to step length of the BTTA group was at most only 1% of the overall step length. For the slow walking speed condition, pelvic contribution was actually negative (-4%). Clinically, it is generally believed that persons with bilateral lower-limb amputations rely to a greater extent on pelvic rotation than able-bodied individuals to step while walking. The data from this study, however, refutes this belief. Although an increase in pelvic rotation in the transverse plane does increase contribution to step length, it appears to play a smaller role than previous studies seem to suggest (Inman et al. 1981; Murray et al. 1966).

This study shows that there are differences in the gait patterns of the BTTA group compared to able-bodied persons. Differences in the contributions by the trailing and leading limbs to step length compared to able-bodied persons may be due to different gait strategies to increase the person's sense of stability, but the data also suggest that changes in segment contributions are due to increased knee flexion of the limbs, which could be related to socket design and/or the alignment of the prostheses.

Chapter 7: Case studies: gait analyses of persons with partial foot amputation walking barefoot and with dorsiflexion stop ankle-foot orthoses

7.1 Introduction

Due to advances in surgery, it has become feasible to perform a transmetatarsal amputation (TMA) on patients as opposed to more proximal levels of amputation. Approximately 10,000 TMAs were performed in the U.S. in 1991 (Mueller and Sinacore 1994). A person with a partial foot amputation (PFA), which retains the calf and ankle musculature, may naturally be expected to walk almost identically to that of an able-bodied person. However, reduced walking speeds (Burnfield et al. 1998; Mueller et al. 1998; Salsich and Mueller 1997; Tang et al. 2004) and step lengths (Burnfield et al. 1998; Dillon 2001) are typically observed for this population. For most levels of PFA, the shortened foot leads to a shorter lever arm for the ankle joint, thereby reducing achievable ankle plantar flexion moments and increasing vertical ground reaction forces on the contralateral limb during loading (Burnfield et al. 1998; Dillon 2001). Those individuals with PFAs also exhibit decreased power generation across the ankle joint compared to able-bodied persons (Dillon 2001; Mueller et al. 1998). Increased power generation at the hip may be necessary to compensate for the loss at the ankle in order to advance the body forward during walking (Dillon and Barker 2006b). These changes in gait may lead to muscle or joint problems over time (Powers et al. 1994; Snyder et al. 1995).

The short residual limb in persons with PFA makes it difficult to create a prosthesis or orthosis that is attached securely and does not inhibit other joint movement or cause discomfort. A device that creates sufficient contact with the residual limb and increases the useable foot length (i.e. increases the effective forefoot rocker) is desirable (Kulkarni et al. 1995). This would potentially allow the person to walk with a gait pattern that is more like that of able-bodied persons. Although many prosthetic and orthotic devices have been described for the management of PFA, their impact on gait has not been evaluated sufficiently (Dillon et al. 2007). A systematic review by Dillon et al. (2007) concluded that PFA affected temporospatial, kinematic, kinetic, and plantar pressure variables during gait, but that there was low or insufficient evidence regarding how these aspects were altered, particularly with the use of assistive devices. Hence, the need still exists to demonstrate the impact of different prosthetic/orthotic devices on the gait of persons with PFA.

The purpose of this study was to examine the impact of a laminated ankle foot orthosis (AFO) with dorsiflexion stop on the gait of persons with PFA. The AFO was designed with three elements hypothesized to be required to increase COP excursion beyond the distal end of the residuum: a rigid toe lever, control of forward progression of the tibia to couple the AFO device with the residual limb, and a substantial socket capable of managing the external torques caused by loading the forefoot (Dillon 2001; Dillon and Barker 2006a; b). It was hypothesized that subjects walking with a device that allows increased COP excursion will display gait characteristics more similar to able-bodied persons. Case studies of two adult males with PFA walking barefoot and with the device are presented in this study.

7.2 Methods

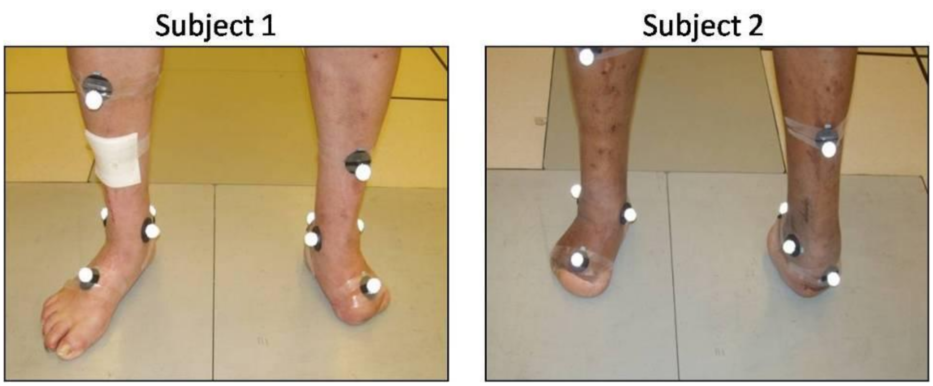

Figure 7.1: Marker placement on the shank and feet of both subjects during barefoot walking.

Two subjects participated in this study as approved by the Northwestern University Institutional Review Board and gave written informed consent. Subject 1 was a 59 year old male (height: 174.5 cm, mass: 98.5 kg) with a short TMA on the left side (Figure 7.1). Subject 2 was a 48 year old male (height: 196.0 cm, mass: 112.5 kg) with a long TMA on the right side and mid-tarsal amputation on the left side. Amputation in both subjects was the result of peripheral vascular disease. Subjects were fitted with an AFO on their left (shorter) side that consisted of a laminated dorsiflexion stop AFO with free plantar flexion, anterior shell, and stiff laminated base with a prosthetic forefoot attached (Figure 7.2). This device was built under the assumption that a device that crosses the ankle may significantly increase anterior translation of the center of pressure and improve functional ability compared to walking without the device (Wening et al. 2008). Subject 2 also walked with a foam toe filler in his right shoe.

Both subjects walked barefoot across a level walkway in the VA Chicago Motion Analysis Research Laboratory (VACMARL) at their freely-selected normal walking speed. An eight-camera Eagle Digital Real-Time motion measurement system (Motion Analysis Corporation, Santa Rosa, CA) was used to acquire marker movements at 120 Hz while ground reaction forces were acquired using six AMTI (Advanced Mechanical Technology, Inc., Watertown, MA) force platforms simultaneously recorded with the motion analysis cameras at 960 Hz. After obtaining at least five clean force platform hits for each leg, subjects donned their athletic shoes and AFO (on their left limb) and again walked at their freely-selected walking speeds.

Figure 7.2: Ankle foot orthosis worn on the left limb by both subjects.

A modified Helen Hayes marker set was used to define a biomechanical model on each person. Markers on the lower limbs were placed on the sacrum, anterior superior iliac spines (ASIS), femoral condyles, malleoli, heels, thigh, shank, and the dorsum of the feet between the fore-foot and mid-foot (or on the shoe at the level where this would be for the intact foot). When walking barefoot, toe markers on the amputated feet were placed on the dorsum of the foot as distal as possible (Figure 7.1). When walking with

Figure 7.3: Front and side views of Subject 1 showing marker placement on feet with shoes and AFO.

AFO and shoes, the ankle markers were placed on the mechanical ankle joint or on the shoe when the anatomical ankle joint was obscured (Figure 7.3). Static standing trials were performed for both the barefoot and AFO walking conditions in order to provide improved estimates of the locations of the ankle joint centers of rotation. Markers were placed on the medial femoral condyles and ankles for the static trial but were removed for the walking trials.

Data from the two PFA subjects were compared to those previously collected from ten able-bodied subjects (mean age: 25 ± 1 years, height: 174.6 ± 9.1 cm, and mass: 69.3 ± 9.5 kg) walking at their freely-selected walking speed. Data were processed using EVa and Orthotrak software (Motion Analysis Corporation), to obtain temporospatial, kinematic, and kinetic information. Data were further processed using custom macros in Microsoft Excel (Microsoft Corporation, Redmond, WA) and Matlab (The Mathworks, Inc, Natick, MA). Temporospatial, kinematic, and kinetic data, as well as center of pressure and ankle-foot roll over shapes were compared between the PFA and able-bodied groups.

7.3 Results

7.3.1 Temporospatial parameters

Subject 1 increased his speed from 0.9 m/s to 1.02 m/s when walking with the AFO device compared to walking barefoot, while Subject 2 increased his speed from 0.80 m/s to 0.83 m/s. Despite these increases, the freely-selected walking speed remained less than that of able-bodied subjects (1.34 m/s). The increase in speed of the PFA subjects walking with the AFO was due to an increase in step length, since each subject's cadence was observed to decrease slightly (Table 7.1). Swing and stance phases were more symmetric bilaterally when walking with the AFO compared to barefoot, with a decrease bilaterally in single limb support and swing time and an increase in double limb support time (as a percent of gait and also as absolute values). Contralateral heel contact occurred for both subjects at approximately 50% of the gait cycle.

Table 7.1: Temporospatial Measurements

	SUBJECT 1				SUBJECT 2				ABLE-BODIED
	Barefoot		**Shoe/AFO**		**Barefoot**		**Shoe/AFO**		**Freely Selected**
Measures	Left	Right	Left	Right	Left	Right	Left	Right	
Speed (m/s)	0.90		1.02		0.80		0.83		1.34
Cadence (step/min)	106		100		105		94		110
Step Length (m)	0.55	0.47	0.64	0.58	0.50	0.41	0.52	0.54	0.73
Single- support time (% gait cycle)	36.5	40.8	35.4	36.0	40.2	36.8	32.3	33.8	39.4
Double-support time (% gait cycle)	12.7	9.8	15.5	13.5	13.6	9.6	18.4	15.5	10.7

7.3.2 **Joint kinematics and kinetics**

Few differences in the joint kinematics or kinetics were observed for either subject walking barefoot and with AFO. There was little to no ankle dorsiflexion observed on the amputated limbs when walking barefoot (Figure 7.4). With the AFO, an increase in ankle dorsiflexion was measured on the left side, comparable to that of the able-bodied subjects. Plantarflexion range was similar to the able-bodied group during stance, but was either lower or did not occur during the swing phase of gait for both the barefoot and AFO conditions.

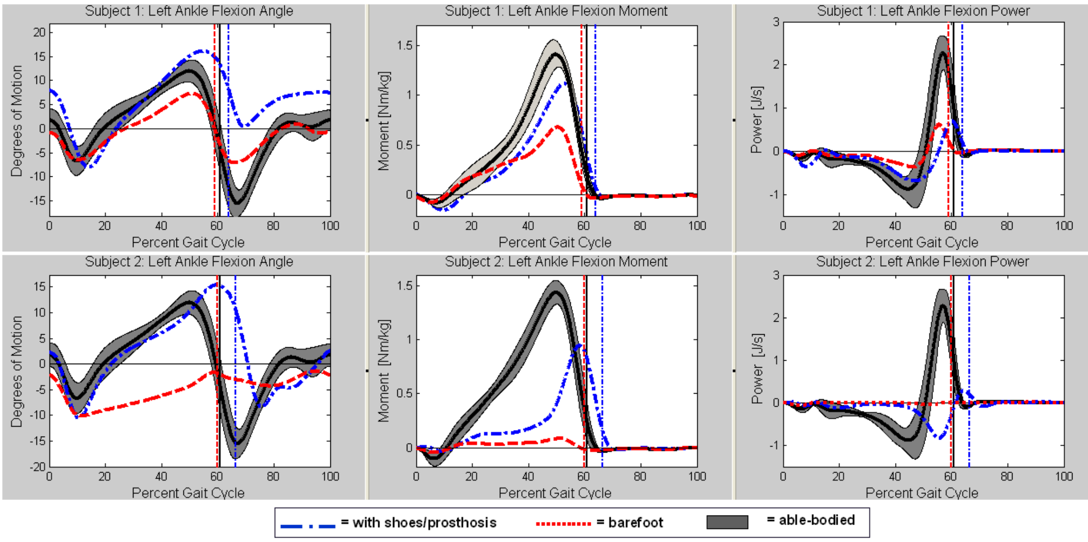

Figure 7.4: Left side ankle flexion angle, moment, and power for Subject 1 (top) and Subject 2 (bottom) walking barefoot and with AFO. Plots are over a gait cycle from heel contact to ipsilateral heel contact. Vertical lines indicate the average time of ipsilateral toe-off. Grey band indicates mean ± one standard deviation for 10 able-bodied subjects walking at their self-selected speed.

The peak internal ankle plantar flexion moment increased when using the AFO compared to barefoot walking, but the peak occurred later in the gait cycle than in able-bodied walking. For Subject 2, the moment was

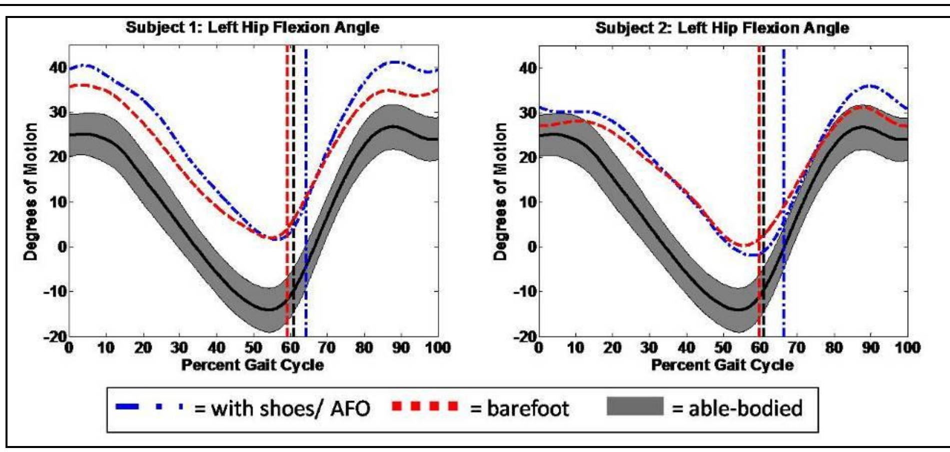

Figure 7.5: Left side hip flexion angle for both subjects walking barefoot and with AFO. Plots are over a gait cycle from heel contact to ipsilateral heel contact. Vertical lines indicate the average time of ipsilateral toe-off. Grey band indicates mean ± one standard deviation for 10 able-bodied subjects walking at their self-selected speed.

negligible until about 50% of the gait cycle and quickly peaked at about 58% of the gait cycle, after contralateral foot contact occurred (Figure 7.4). Sagittal plane ankle joint powers were much lower on both subjects' left (shorter) sides, with no appreciable increase in joint power observed while walking with the shoe and AFO. Additionally, peak power absorption and generation were observed to occur later in the gait cycle compared to able-bodied walking. On the right limbs of both subjects, there was little or no difference between walking with the AFO or barefoot. Lower ankle plantar flexion moment and power peaks were observed compared to those observed in able-bodied subjects on the right sides.

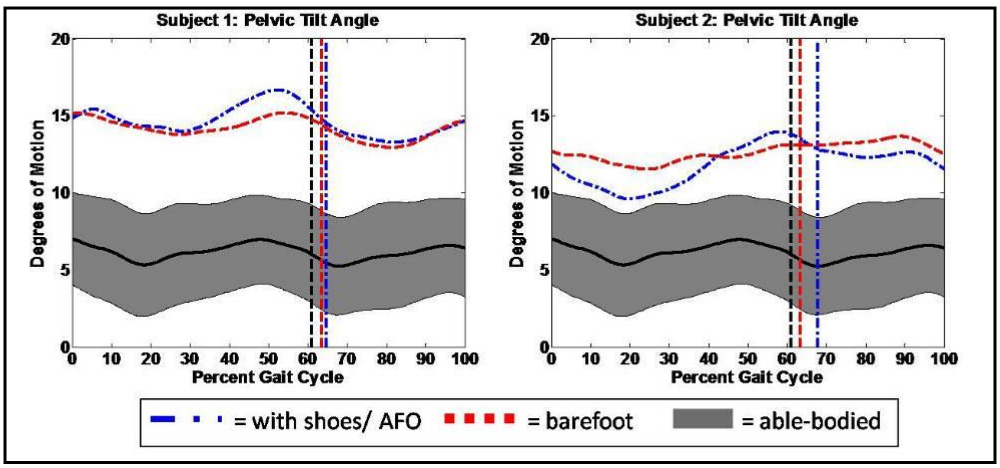

Figure 7.6: Pelvic tilt angle from left heel contact to subsequent left heel contact for both subjects walking barefoot and with AFO. Vertical lines indicate the average time of ipsilateral toe-off. Grey band indicates mean ± one standard deviation for 10 able-bodied subjects walking at their self-selected speed.

An offset was observed in hip flexion angle compared to that of able-bodied data (Figure 7.5), with increased hip flexion and decreased hip extension in the PFA subjects. There was very little if any difference in either subject's hip flexion curves between walking barefoot or with the AFO. Subject 1 displayed slightly increased hip flexion peaks compared to walking barefoot, but no difference in hip extension was observed. An increase in the pelvic tilt offset compared to able-bodied walking was also observed for both barefoot and AFO walking conditions, ranging from 10 to 16° over the gait cycle (able-bodied range: 2° to 10°) (Figure 7.6).

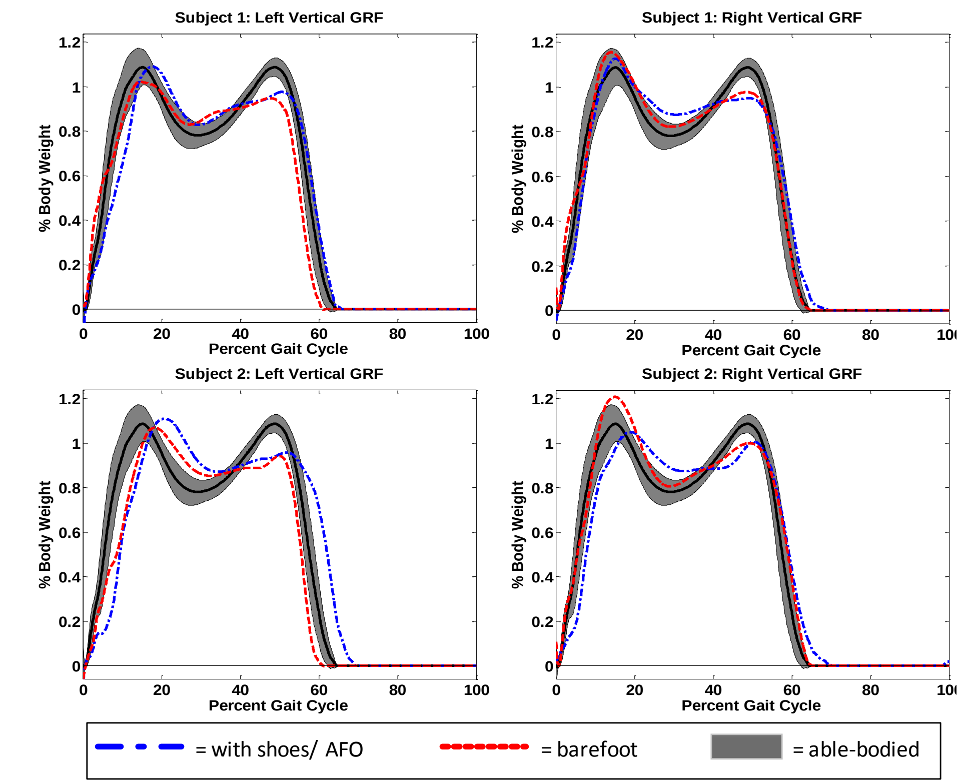

Figure 7.7: Vertical ground reaction force for both subjects' left and right sides walking barefoot and with AFO.

7.3.3 **Pressure and force measurements**

The magnitudes of the first peak of the vertical ground reaction forces (vGRF) were similar to those observed in able-bodied persons, but the timing was delayed on the left side when walking with the AFO for Subject 1 and on both left and right sides for Subject 2 (Figure 7.7). On both subjects' right side, first peak of the vGRF was higher during barefoot walking compared to the able-bodied data and also compared to walking with shoes and the AFO device. The second peak of the vGRF was lower

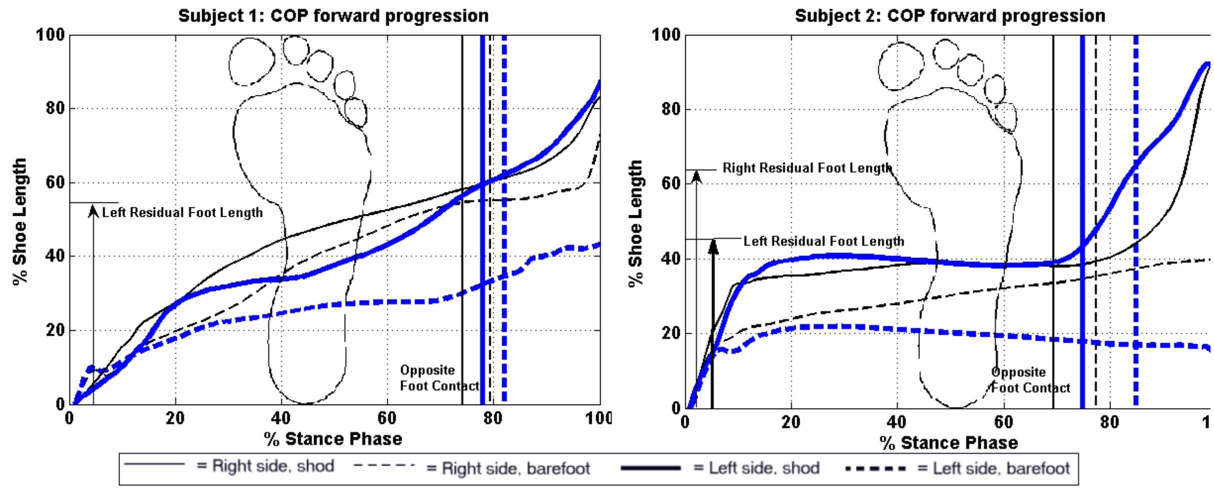

Figure 7.8: Center of pressure (COP) forward progression (measured as a percentage of total shoe length) over the stance phase of gait for both subjects. Vertical lines denote when contralateral foot contact occurred. COP progression patterns became more symmetric between left (thick lines) and right (thin lines) sides when walking with the shoe and AFO (solid lines) compared to barefoot walking (dotted lines).

bilaterally for the two subjects when walking both barefoot and with AFO compared to that observed in able-bodied subjects. Though the magnitudes of the second peaks were similar between barefoot and AFO walking conditions, drop-off of the vGRF occurred earlier in the gait cycle for barefoot walking compared to walking with the AFO.

When walking barefoot, the center of pressure (COP) under the left foot of both subjects progressed from the heel along half of the residual foot length before contralateral foot contact occurred (Figure 7.8). Anterior translation of the COP increased under the left foot when using the AFO and shoe, and COP progression patterns between left and right sides were more similar compared to walking barefoot. With the AFO on the left foot, COP progressed close to the end of the residual foot length, but did not move beyond the residual foot length until after contralateral foot contact. For Subject 2's right side, in which a foam toe filler was used, the COP was only able to progress about two-thirds of the total residual foot length before contralateral foot contact occurred.

Ankle-foot roll over shapes (Hansen et al. 2000) were also calculated for both subjects and were compared to that of the able-bodied group (Figure 7.9). These shapes are the effective rocker shapes that the ankle-foot system conforms to between heel contact and opposite heel contact and are determined by converting the center of pressure from a world-based coordinate system to a shank-based coordinate system. The effective forefoot length (EFL), measured as the sagittal plane distance from the ankle marker to the anterior position of the roll over shape, was calculated for each subject (Table 7.2). Both subjects' EFL increased about 5 cm on their left side when walking with the AFO device versus walking barefoot. Little change was noted on their right sides. Though the EFL of the PFAs was less than that of the able-bodied ambulators in all cases, left

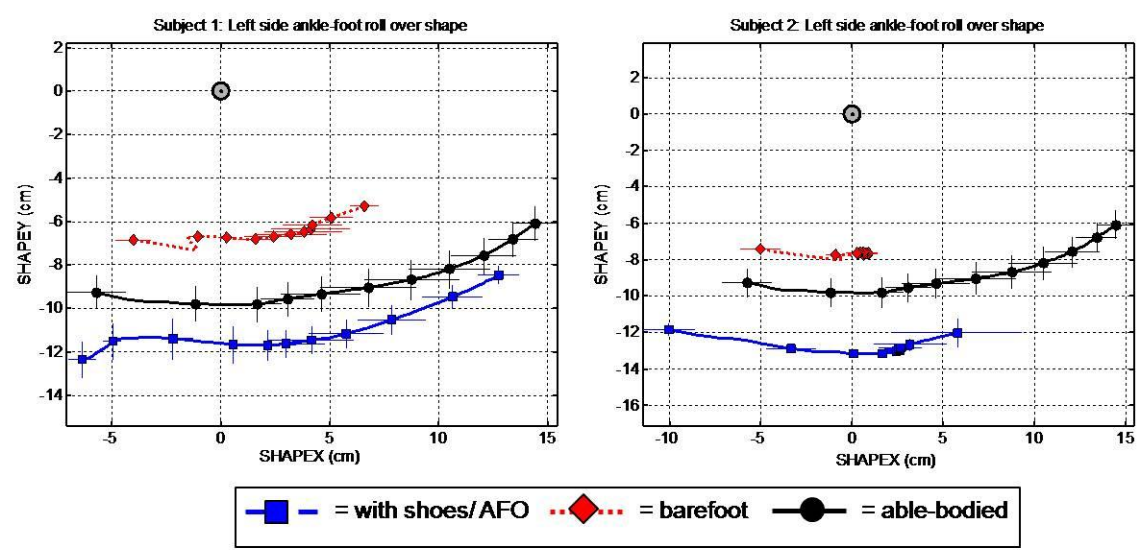

Figure 7.9: Left side ankle-foot roll over shapes for both subjects walking barefoot and with AFO. Mean able-bodied roll over shape for self-selected walking is included on both graphs for comparison.

Table 7.2: Effective forefoot length

	SUBJECT 1				SUBJECT 2				ABLE-BODIED
	Barefoot		**Shoe/AFO**		**Barefoot**		**Shoe/AFO**		**Freely Selected**
Measures	Left	Right	Left	Right	Left	Right	Left	Right	
Effective forefoot length (cm)	6.6	12.6	12.8	13.7	1.0	6.4	5.9	6.3	14.7

and right sides were more symmetrical with the AFO device compared to walking barefoot.

7.4 Discussion

The walking speeds of the two PFAs were similar to those previously reported (0.84 m) for PFAs with a history of diabetes or vascular disease (Dillon et al. 2007). The lower speed was due to both a reduction in cadence and step length compared to the able-bodied group. An increase in step length was observed when walking with shoes and the AFO device compared to barefoot walking with as much as 13 cm increase in step length (observed on Subject 2's right side step length). This is similar to results found by Wening et al. (2008) who fit one subject with a similar device. It is unclear how much of the increase can be attributed to the AFO because an increase in step length also occurred for Subject 1 on the intact limb between barefoot walking and walking with shoes. A study performed by Tang et al. (2004) saw increases in walking speed between barefoot walking and walking with shoes (no AFO device) both for persons with transmetatarsal amputation and able-bodied persons, though these differences were not found to be significant.

Despite modifying the Helen Hayes model and placing the toe marker over the distal end of the residua, substantial dorsiflexion was measured in a device that was designed to prevent that specific motion to a large extent. Although this marker modification would likely have eliminated ankle motion measurement errors due to forefoot bending, it would not have eliminated errors due to heel slippage (Dillon et al. 2008). Hence, given that the markers that defined the shank and foot were distributed across the anatomical limb, the AFO, and the shoe, possible sources of error in these case studies include relative movement between the shank and AFO as well as relative motion between the device and residua with respect to shoe.

The AFO was designed with elements that were hypothesized to increase COP excursion beyond the distal end of the residuum: a rigid toe lever, coupling between the residual limb and device to control of forward progression of the tibia (dorsiflexion stop), and a substantial socket (anterior shell) capable of managing the external torques caused by loading the forefoot (Dillon 2001; Dillon and Barker 2006a; b). Although the AFO increased COP excursion anteriorly compared to barefoot (and therefore created a longer effective forefoot length), the COP did not move beyond the distal end of the residuum in either subject until contralateral initial contact occurred. This result concurs with the drop-off observed in the second peak of the vGRF for both subjects and increased loading on the leading limb (first peak of the vGRF), which is indicative of an inability to load the forefoot effectively (Hansen et al. 2006). There were slight differences in ankle motion and moment between the two subjects at the time of contralateral initial contact, suggesting that either the dorsiflexion stop engaged at different times or that there is potentially a relationship between residual limb length and time at which the dorsiflexion stop needs to engage in order to improve COP excursion.

It has been suggested that this type of device may not be appropriate for lower-level ambulators who take short steps or have loss of sensation due to diabetes (Wening et al. 2008), possibly due to the inability to load the residual limb or allow progression of the COP onto the orthosis. It is possible then that these subjects walked too slow or were not capable of bearing the load required on their residual limb to fully utilize the AFO device. A study of persons with PFA capable of walking at faster freely-selected speeds may produce different results. It has been shown that other devices such as the clamshell prosthesis are able to allow the COP to progress past the residual limb (Dillon 2001). A study to determine differences between these devices, and whether changes in alignment or fit of the AFO can increase the anterior excursion of the COP requires further exploration.

Chapter 8: Segment contribution to step length for persons with partial foot amputation

This chapter is added as an addendum to Chapter 7.

Persons with partial foot amputation (PFA) walk with different gait characteristics than able-bodied persons, possibly due to their prosthesis, feeling of instability, or inability to load the forefoot during walking. This addendum to Chapter 7 looks at the lower limb segment contributions to step length for two persons with partial foot amputation (one unilateral, one bilateral). The percentage contributions of each segment to overall step length were compared to those of able-bodied persons walking at their freely selected walking speed and step length and also at a speed where step lengths were similar (Table 8.1). The SCSL analysis calculates the contribution of six lower limb segments to the overall step length. The segments that were included were the trailing ankle-foot, shank, and thigh segments, the pelvic segment, and the leading thigh and shank segments. This type of analysis will enable us to determine how differences in step length during gait occur between able-bodied persons and persons with amputation.

Table 8.1: Temporospatial Measurements

	SUBJECT 1				SUBJECT 2				ABLE-BODIED	
	Barefoot		**Shoe/AFO**		**Barefoot**		**Shoe/AFO**		**Freely Selected**	**Matched**
Measures	Left	Right	Left	Right	Left	Right	Left	Right		
Speed (m/s)	0.90		1.02		0.80		0.83		1.34	1.14
Cadence (step/min)	106		100		105		94		110	114
Step Length (m)	0.55	0.47	0.64	0.58	0.50	0.41	0.52	0.54	0.73	0.66

Additionally, the SCSL analysis will indicate how step length is affected with an AFO device designed to increase the effective forefoot length and the center of pressure progression compared to when these subjects walk barefoot.

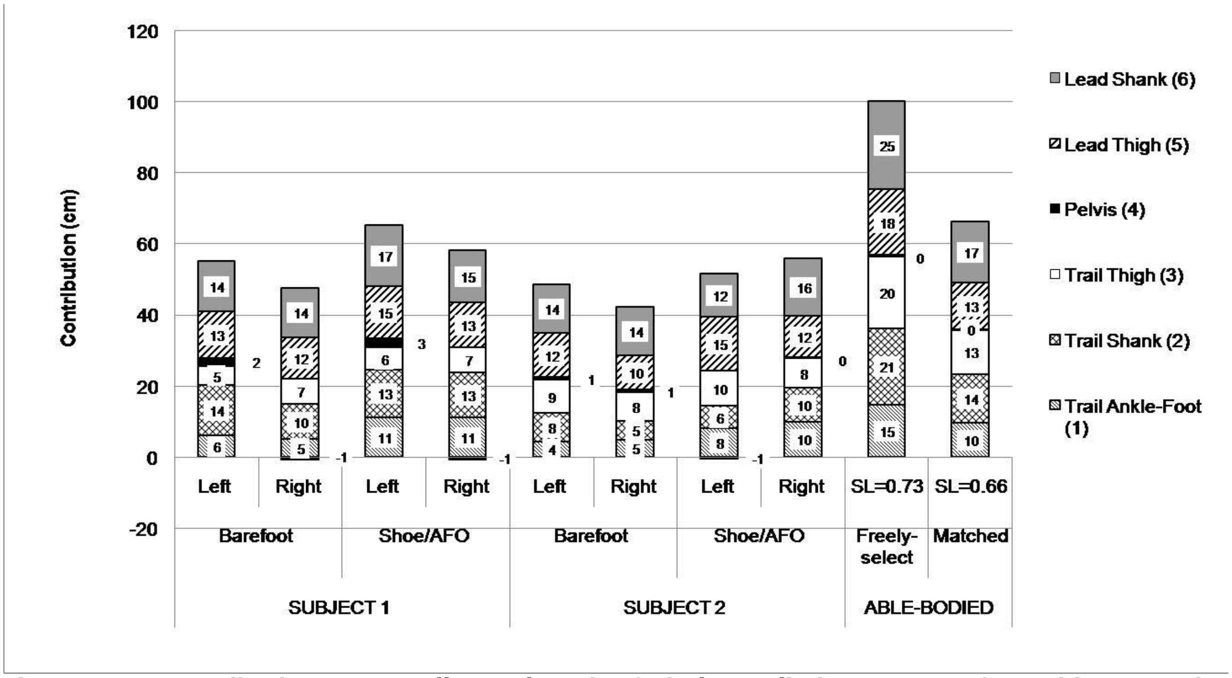

Figure 8.1: Contribution to overall step length of six lower limb segments for Subject 1 and 2 walking barefoot and with AFO at freely-selected walking speeds. Left and right sides are reported separately for these subjects. Mean segment contribution to step length for the able-bodied group are also reported for freely-selected walking and walking at similar step lengths to the PFA subjects.

A graph of the each segment's contribution to overall step length for the left and right limbs of each PFA subject walking barefoot and with AFO is plotted in Figure 8.1. Contributions as a percent of overall step length is reported in Figure 8.2. Comparisons were made between barefoot and AFO conditions of each subject. Comparisons were also performed between the PFA subjects and able-bodied subjects when able-bodied persons walked at a similar step length. Percent contributions of the segments were similar for the able-bodied group when walking at freely-selected (1.35 m/s) and matched (1.14 m/s) walking speeds.

8.1 Barefoot and AFO comparisons of persons with partial foot amputation

8.1.1 Subject 1 with Unilateral PFA

Subject 1's left side step length increased by 11 cm when walking with the AFO compared to walking barefoot. Main contributors to the increased step length were observed by the trailing ankle-foot (3 cm), leading thigh (2 cm) and leading shank (3 cm) segments. As a percent contribution, increased percent contribution was only observed by the trailing ankle-foot segment (6%), while decreases occurred for the trailing shank (5%) and leading thigh (2%) segments.

On the right (intact) side, step length also increased by 11 cm when walking with the shoe compared to walking barefoot. Increased segment contribution was observed in the trailing limb for both the ankle-foot (6 cm) and shank segments (3 cm). Increases in the leading limb of only 2 cm was observed. The largest contributor as a percentage contribution occurred by the trailing ankle-foot segment (8%), with only a 1% increase by the trailing shank. The trailing thigh and leading limb segments decreased by approximately 3% of overall step length.

8.1.2 **Subject 2 with Bilateral PFA**

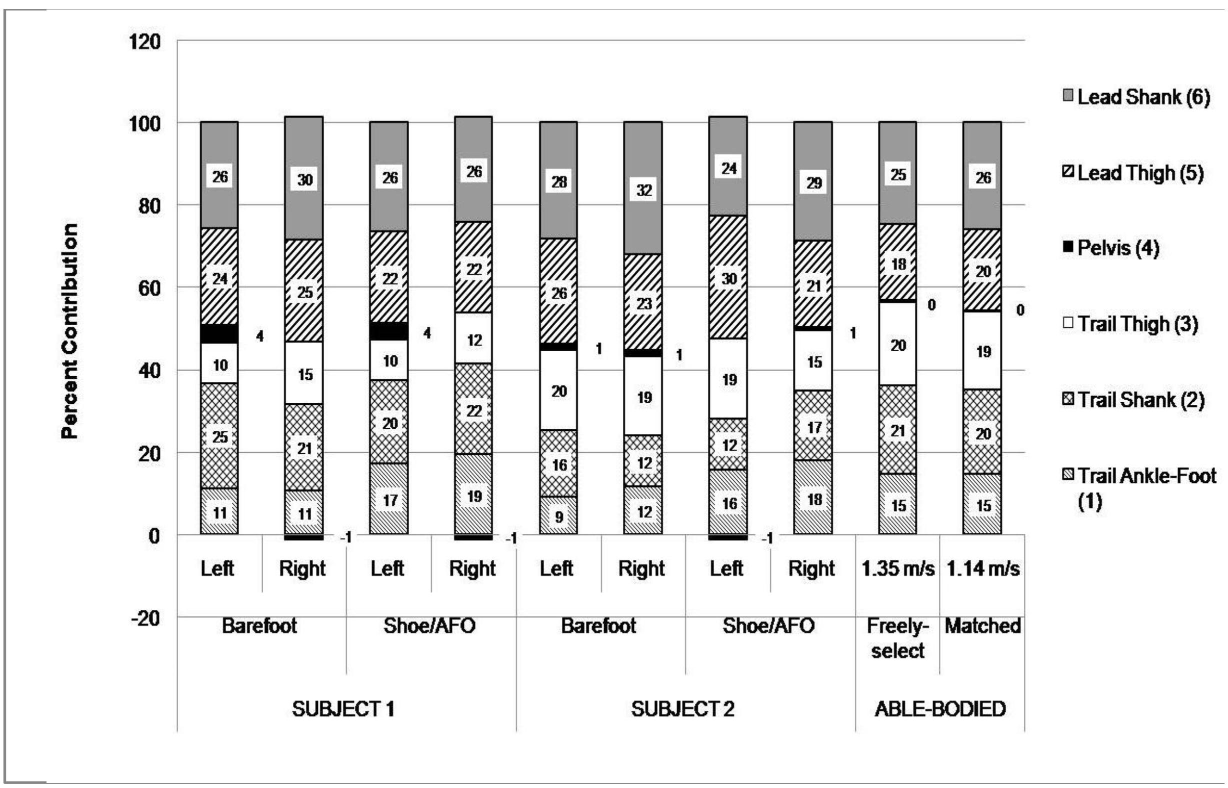

Figure 8.2: Percent contribution to overall step length of six lower limb segments for Subject 1 and 2 walking barefoot and with AFO at freely-selected walking speeds. Left and right sides are reported separately for these subjects. Mean segment contribution to step length for the able-bodied group are also reported for freely-selected walking and walking at similar step lengths to the PFA subjects.

Compared to barefoot walking, Subject 2’s left side step length only increased by 2 cm when walking with the AFO. Even though only a small increase was observed in step length, larger changes of each contribution were observed. Increases in contribution to step length on the left side were observed in the trailing ankle-foot segment (4 cm) and thigh (1 cm), while decreases were observed by the trailing shank (2 cm), and leading thigh (3 cm) and leading shank (2 cm). As a percent contribution to overall step length, increases of 5% were observed for the trailing ankle-foot segment and 4% by the leading thigh segment. Decreases of 4% for the trailing and leading shank segments and 1% by the trailing thigh and pelvic segments occurred when walking with the AFO.

On the right side, an increase in step length of 14 cm was observed, attributable to increases in contributions by the trailing ankle-foot and shank segments of 5 cm each and by the leading thigh and shank segments of 2 cm each. As a percent contribution, increases were observed by the trailing ankle-foot (6%) and trailing shank (5%) segments while decreased contributions were noted by the trailing thigh (4%), leading thigh (2%), and leading shank (3%)

8.2 Comparisons between able-bodied persons and persons with PFA

Comparisons were also performed between the percent contributions of each segment for the PFA subjects and able-bodied subjects when they walked at step lengths that were similar. The shortest step length acquired from the able-bodied subjects was 0.66 m, which was still greater than those observed in the PFA subjects by 2-25 cm.

Comparisons were performed between the able-bodied data walking with shoes and with the PFA subjects walking barefoot and with their AFOs.

8.2.1 Comparison between PFA subjects walking barefoot compared with able-bodied persons

On the left side, Subject 1 had lower contributions by the trailing ankle-foot (4%) and thigh (9%), but increased contributions by the trailing shank (4%), pelvis (4%), and leading thigh (6%). On the right side, the segments with decreased contributions were similar to those on the left, with decreases observed for the trailing ankle-foot (4%) and thigh (4%) segments. Increased contributions were observed by the leading thigh (5%) and shank (4%).

For Subject 2, the segmental contributions on the left side were lower compared to the able-bodied group for the trailing ankle-foot (6%) and shank (4%), but were greater for the pelvis (1%), leading thigh (6%) and leading shank (2%). Similar to the left side, decreases in contributions occurred on the trailing limb on the right side while increases occurred on the leading limb. Percent contributions were 3% lower for the trailing ankle-foot and 8% less for the trailing shank, while they were 1%, 3% and 6% greater for the pelvis, leading thigh and shank, respectively.

8.2.2 Comparison between PFA subjects walking with AFO and able-bodied persons

With the AFO, Subject 1 had greater contributions on the left side by the trailing ankle-foot (2%), pelvis (4%) and leading thigh (2%) segments compared to able-bodied walking. A lower contribution was only observed in the trailing thigh (9%). On the right side, greater contributions of 4% for the trailing ankle-foot, 2% by the trailing shank, and 2% by the leading thigh were observed compared to the able-bodied group, while lower contributions were observed by the trailing thigh (7%) and pelvis (1%).

Subject 2 had slightly increased contributions on his left side by the trailing ankle-foot (1%), but a large increase in lead thigh contribution (10%) compared to the able-bodied group. Lower contributions occurred for the trailing shank (8%), pelvis (1%), and leading shank (2%). On the right side, Subject 2 had greater contributions for the stance ankle-foot (3%), pelvis (1%), lead thigh (1%), and lead shank (3%), and lower contributions by the trailing shank (3%) and thigh (4%).

8.3 Conclusions

When walking with the AFO, Subject 1 increased walking speed by 0.12 m/s compared to barefoot walking and increased step length on both limbs by 11 cm . Contributions by the ankle-foot segment played a large role in increasing step length, which was achieved by increased anterior progression of the center of pressure (COP) under the foot. Wearing the AFO device may have increased the subject's sense of stability, since the leading limb contributions increased (swing limb progressed out further) for both the sound and impaired legs. Increased contributions of the ankle-foot segment for the right limb (with the intact foot) when walking with shoes was also observed, so it is difficult to determine how much the AFO actually helped compared to that of a shoe itself.

Subject 2 demonstrated a large increase in step length on the right side when walking with AFO compared to barefoot walking, an effect that is believed to be primarily due to the AFO. Increases in the segment contributions occurred primarily in the trailing (left) limb at the level of the ankle-foot and shank, which implies that the COP progressed further anteriorly and created a longer lower limb rocker arc as described by Perry (1992). It was believed that the small change in step length of only 2 cm on the left side occurred primarily because the subject walked with a foam toe filler on his right side, which was not sufficiently stiff to allow loading beyond the most distal aspect of the residual limb. This occurs because the ability to load the contralateral limb during late stance is a determinant to the swing distance of the ipsilateral limb (and thus affects step length. Therefore, it would then be assumed that few gait changes would occur between walking barefoot and with shoes having a foam toe filler. However, changes occurred in all segments, with increased contributions by the trailing foot segment and thigh, and decreased contributions by the trailing shank and leading limb (shank and thigh segments). It is possible that a “drop-off” effect on the trailing limb occurred, decreasing the ability of the leading leg to swing forward. Though slight increases in segmental contributions were observed to occur for the trailing leg, this was offset by reduced contributions in the leading leg.

When comparing the step length segmental contributions of the PFA subjects with the able-bodied group, lower percent contributions by the trailing ankle-foot segments during barefoot walking were observed. This implies that walking with the shorter foot (i.e., rocker arc) length decreases the ability of the trailing limb to contribute to the overall step leading step length. A greater pelvic contribution was observed in the PFA subjects compared to the able-bodied group, though this was only by 4% of overall step length for Subject 1 and 1% for Subject 2. Contributions by the other segments varied between Subjects 1 and 2, so general conclusions about how they compare with the able-bodied group could not be made for these segments. For Subject 2, a lower percent contribution by the trailing limb and an increased contribution by the leading limb for both the left and right sides were observed when walking barefoot compared to the able-bodied group. The short residual foot lengths of both limbs prevented a COP progression pattern that was similar to that of the able-bodied group.

When walking with the AFO, the PFA subjects' segment contributions were more similar to that of the able-bodied group than when they walked barefoot. Larger percent contributions by the trailing ankle-foot and shank segments, and decreased percent contribution by the leading limb were observed. Even while walking with the AFO, differences still existed between the PFA subjects and able-bodied group, with the larger differences occurring in the trailing thigh and pelvis for Subject 1 and in the trailing shank and leading shank and thigh for Subject 2. An able-bodied population that was better matched to the step length of each PFA subject may provide additional insight into the different means of achieving step length by the PFA population. Better conclusions could also be drawn with a larger group of persons with partial foot amputation, particularly one that was more homogeneous (e.g. all persons having a unilateral amputation at the same level).

Chapter 9: Concluding remarks

These studies explored the methods by which different groups of people modulated their step length. It was hypothesized that step length is modulated by several different means: increasing hip flexion and extension, increasing ankle-foot roll over arc length; increasing stance foot heel rise (by sagittal rotation about the ball of the foot) to further extend the trailing limb, and increasing pelvic rotation. Gait analyses were performed for able-bodied persons, race walkers, persons with bilateral trans-tibial amputation and persons with partial foot amputation. Previous research has shown that freely-selected walking speeds of persons with amputation are slower than that of able-bodied persons, which is at least partially attributable to a decrease in step length (Dillon 2001; Isakov et al. 1997; James and Oberg 1973; Macfarlane et al. 1997; Mueller et al. 1998; Ruhe 2004; Su et al. 2007; 2008; Tang et al. 2004; Underwood et al. 2004). Step lengths of the persons with amputation in our study were found to be shorter than those of the able-bodied group during freely-selected walking, though they were not significantly different for the persons with bilateral trans-tibial amputation (BTTA) when data were speed matched (Chapter 6).

The Segment Contribution to Step Length (SCSL) analysis was introduced as a method to examine how each of the lower limb segments contributes to the overall step length and how these contributions vary as changes are made in one’s walking pattern (e.g. taking longer steps or after gait training). The SCSL analysis was also used to compare the differences in segment contributions between different subject groups. The six segments contributing to the overall step length were the trailing ankle-foot, shank, and thigh segments; the pelvic segment; and the leading thigh and shank segments. Normative data using able-bodied persons walking at their freely-selected walking speed were reported in Chapter 2. A SCSL analysis was performed on able-bodied persons for increasing step length in Chapter 3 to determine what segments contribute to increased step lengths. Chapter 4 reported the differences in segment contributions of trained race walkers compared to freely-selected and fast walking, while differences in step length contributions between able-bodied persons and persons with bilateral trans-tibial amputation were reported in Chapters 5 and 6 and persons with partial foot amputation in Chapters 7 and 8. Some of the results for each of the subject groups are plotted in Figure 9.1 and Figure 9.2 for comparison. Able-bodied persons walking at short, freely-selected, and longest possible step length are reported, as well as data collected from trained able-bodied race walkers. The short step length was used for comparisons to the groups with lower limb amputation. No significant differences were observed between segments for the BTTA group between the different foot types, so plots for the BTTA subjects wearing the unaltered Shape&Roll foot (long foot condition) is presented. Data from the two subjects with partial foot amputation walking at their freely-selected walking speeds barefoot and with AFO device are also included.

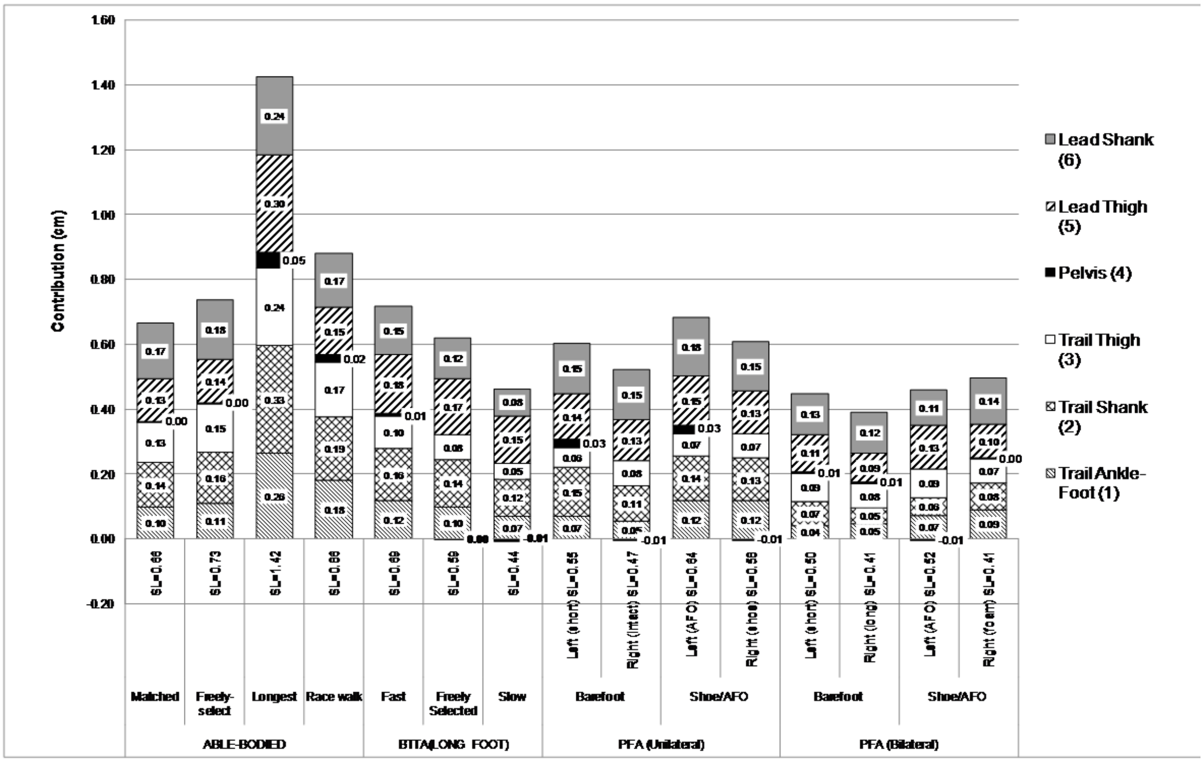

Figure 9.1: Contribution of each lower limb segment to overall step length (normalized by leg length) for able-bodied persons, persons with bilateral trans-tibial amputation (BTTA) walking with Shape&Roll feet, and persons with Partial Foot Amputation (PFA) walking both barefoot and with AFO.

For able-bodied individuals, the trailing ankle-foot segment contributes about 0.1 LL, or 15% to overall step length during freely-selected walking. This contribution increased to 0.18 LL in race walkers and 0.26 LL for longest step lengths of able-bodied persons, though this remained about 16% of overall step length contribution. In contrast, the contribution by the trailing ankle-foot in the persons with partial foot amputation was between 0.04 – 0.07 LL (9%-11%) for barefoot walking. These results, along with those from the analysis of the ankle-foot roll over shapes, implies that persons with normal effective foot lengths (i.e. intact feet) are able to utilize the ankle-foot segment to

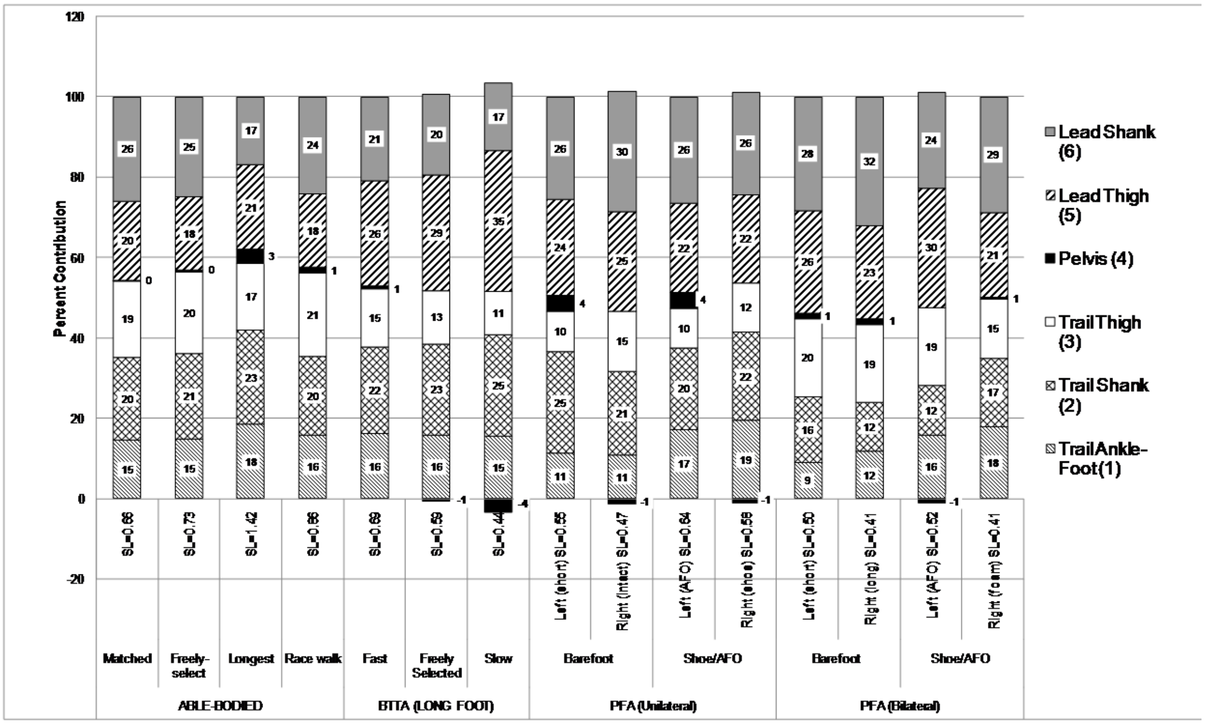

Figure 9.2: Percent contribution of each lower limb segment to overall step length for able-bodied persons, persons with BTTA walking with Shape&Roll feet, and persons with PFA walking both barefoot and with AFO.

increase step length by increasing the ankle-foot roll over arc length as evidenced by the COP to progress further anteriorly onto the forefoot. Additionally, able-bodied individuals are able to increase stance foot heel rise by rolling forward further on the roll over shape arc, while the increased ankle plantarflexion allows active “push-off” of the forefoot, particularly for long steps. These effects also have an influence on the increased contribution of the shank segment for increased step lengths.

Persons having short effective foot lengths will have lower contributions to step length by the trailing ankle-foot segment compared to matched step lengths of able-bodied persons. This was observed in persons with partial foot amputation, and for the BTTA subjects walking at fast speeds with the Short arc length feet. Results of the BTTA subjects show that for increases in speed and step length, increases in the contributions of all segments occurred, and contributions by the trailing ankle-foot and shank were similar to those of the able-bodied group when walking fast. This implies that the unaltered Shape&Roll prosthetic foot used in the study allowed a similar COP progression pattern to that of able-bodied persons during normal walking speeds and step lengths. Decreases in contribution of the trailing thigh and leading shank for the BTTA group relative to able-bodied walking may be due to socket fit, as discussed in Chapter 6.

As expected, the largest contributions to step length were from the shank and thigh segments. For able-bodied persons, percent contribution was generally higher for the trailing thigh segment, while percent contribution by the leading thigh segment was generally lower compared to the subjects with amputation. The persons with BTTA walked with approximately 10° more hip flexion compared to the able-bodied group and had little to no hip extension (Figure 9.3). Similar curves to that of the BTTA group were also displayed by the PFA subjects (Chapter 7). These data imply that persons with amputation walk with different gait strategies than able-bodied individuals, using their leading limbs for step length modulation more than that of able-bodied persons (Figure 9.4). The leading thigh segment had the largest overall contribution to step length for the BTTA subjects due to the increased hip flexion utilized while walking. On the other hand, the trailing limb of the able-bodied subjects contributed a higher percentage (56%) than the subjects with amputation (trailing limb ranges between 43% and 54%).

The ability to remain on the trailing limb for a longer period during the gait cycle and progress the COP further anteriorly under the foot may be the reason for the increased trailing limb contribution for the able-bodied group.

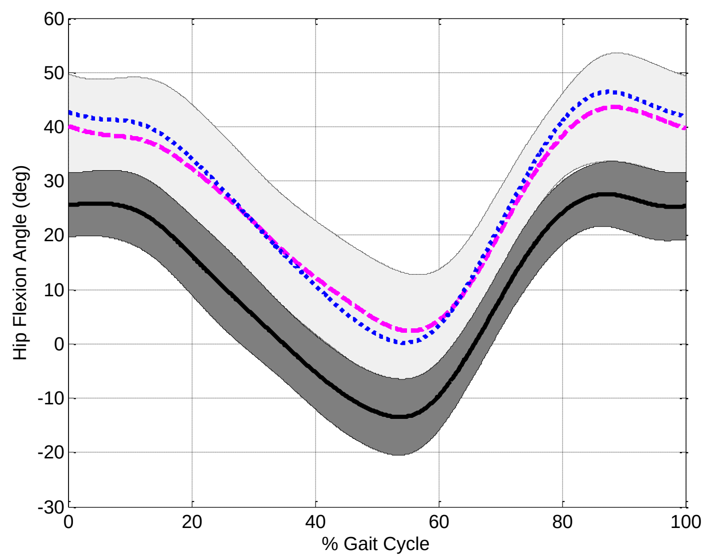

Figure 9.3: Mean hip flexion angles for able-bodied persons (solid line) and persons with bilateral transtibial amputation (BTTA) at freely-selected walking (long dashed red line) and fast walking (short dotted blue line) speeds. Differences in speed and step length were not significant between freely-selected walking of able-bodied subjects and fast walking of BTTA subjects. Shaded areas around the mean line represent one standard deviation for the subject groups. Persons with partial foot amputation displayed similar hip flexion angles to the BTTA group.

Pelvic rotation was expected to play a more significant role to step length contribution than was observed, particularly for race walkers and persons with amputation. However, the pelvis was determined to be the lowest contributor to step length for all subject groups under all walking conditions, with maximum contributions of only 4%. In some cases, pelvic rotation occurred in a manner such that contribution by the pelvic segment was actually negative; that is, instead serving to increase step length, it decreased it. Clinically, it is generally believed that persons with lower-limb amputation rely to a greater extent on pelvic rotation than able-bodied individuals to step while walking. The data from this study, however, refutes this belief. Although an increase in pelvic rotation in the transverse plane does increase contribution to step length, it appears to play a smaller role than previous studies seem to suggest (Inman et al. 1981; Murray et al. 1966). Nonetheless, pelvic rotation may have a larger potential role in progressing the leading limb forward or reducing step width, as suggested by race walking coaches (McGovern 1998; 2005; Salvage 2005).

Further research using the SCSL analysis will allow us to better understand how persons with amputation walk and to determine if decreases in step length observed in

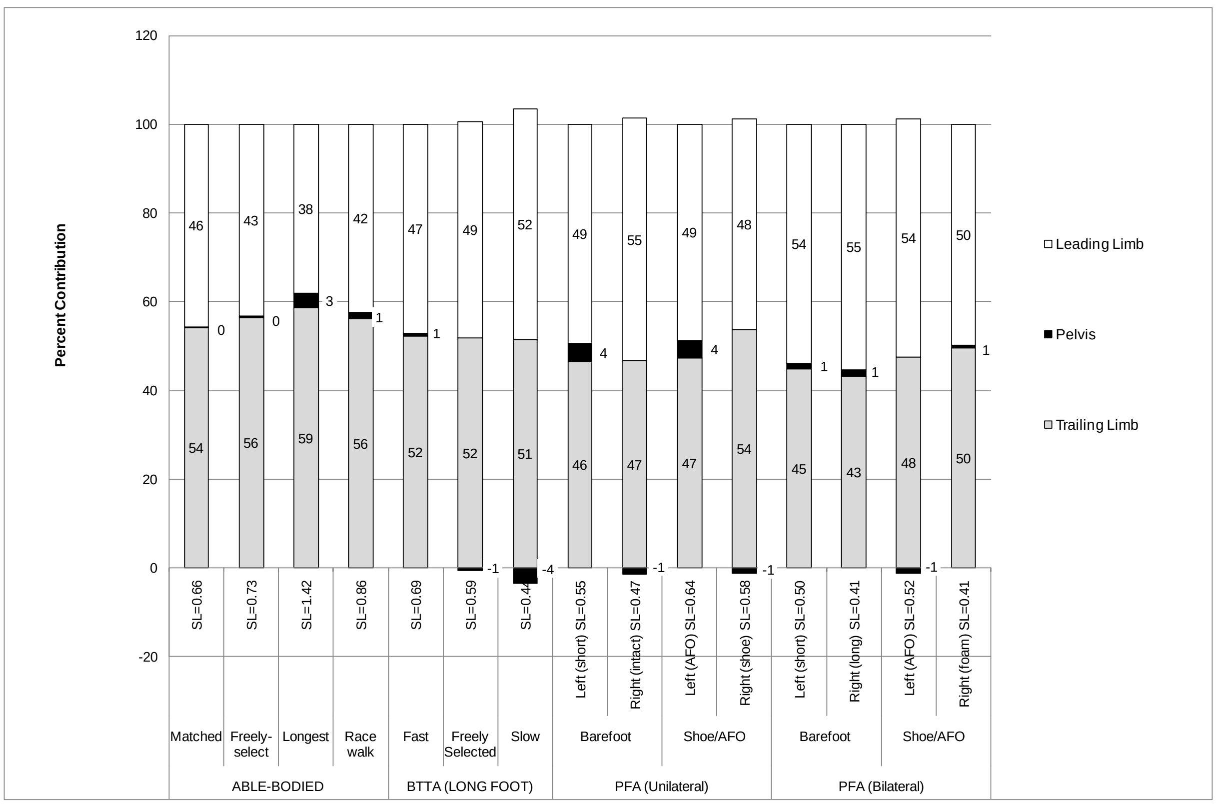

Figure 9.4: Percent contribution by the trailing limb, pelvis, and leading limb for able-bodied persons, persons with BTTA walking with Shape&Roll feet, and persons with PFA walking both barefoot and with AFO. The trailing limb consists of contributions made by the trailing ankle-foot, shank, and thigh segments. The leading limb consists of contributions made by the leading shank and thigh segments.

this population are due to the performance of the prosthesis, fit of the socket, or adaptations to gait in the other lower limb joints. A better cohort of persons with partial foot amputation is needed to better understand general gait characteristics of this population and to evaluate the extent that various interventions have on their gait performance, particularly step length. Further research should include a study of PFA subjects walking barefoot, with shoes only, and with a prosthetic or orthotic device that has the characteristics thought to improve COP progression. SCSL analyses should also be run on other subject populations with lower limb amputation, such as persons with unilateral trans-tibial amputation, as well as persons with unilateral and bilateral trans-femoral amputation. With these results, we can better determine what lower limb segments contribute to an individual’s step, and why these differences occur. This will help to determine if future research should be focused on improving function of the prosthesis and socket or if more focused gait training is needed.

The SCSL analysis is a simple tool that is useful for determining causes of step length differences between subjects and groups of subjects. Reported measurements using the SCSL analysis combine segment length and orientation into one measurement. By knowing where step length differences occur for persons with pathology, we can better determine what training or treatment can improve upon gait. If gait training is performed, a SCSL analysis can help us to determine the specific regions where deficiencies exist, focus training to these specific areas, and help provide objective measure to determine where improvements occur.

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Appendix: Rocker Foot and Roll Over Shapes

We have tried to correlate rocker foot models with experimental data. Using spatial changes of body markers and force plate data, we can create "rocker foot" shapes. The initial idea for this measurement came from Stein and Flowers (1987) who used the invariant relationship of center of pressure (COP) in the direction of forward progression and shank angle to measure the shape of a solid-ankle cushioned heel (SACH) prosthetic foot, and was able to relate it to a rigid cam shape (Figure A.1).

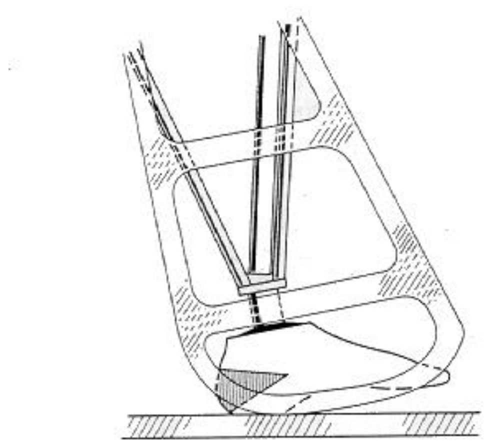


Figure A.1: Foot shape of prosthetic SACH foot overlaid with profile of a rigid cam as depicted by Stein and Flowers (1987).

Knox (1996) later developed similar methods to measure the effective cam shapes of prosthetic feet of non-disabled ankle-foot systems. Hansen (2002) concluded that these shapes could be measured using hip, knee, ankle, and foot markers and it appears that the mechanical characteristics of the limb adapts to changes in weight, heel height, and inclines to keep these shapes the same. These shapes are defined as *roll over shapes* since they are the shapes formed during the gait period from heel contact (HC) to opposite heel contact (OHC), the time the lower extremity is in the "roll-over" phase of gait. Between OHC and toe-off, other goals likely exist for the trailing limb, which is being rapidly unloaded and ceases to act as a rocker (Hansen et al. 2000). The rollover shapes are obtained by transforming the coordinates of the COP along the ground from a world-based coordinate system to a marker-based coordinate system. Three methods to calculate rollover shapes are utilized to include the effects of various lower limb joints, and differ based on which marker-based coordinate system is used. The foot (F) roll-over shape accounts only for the foot; the ankle-foot (AF) roll-over shape includes the effect of the ankle; and the knee-ankle-foot (KAF) roll-over shape includes the effects of the foot, ankle, and knee (Figure A.2). Studies in this dissertation will focus on the roll over shapes created by the foot and ankle (AF roll over shapes). Examples of these transformations are shown in Figure A.4.

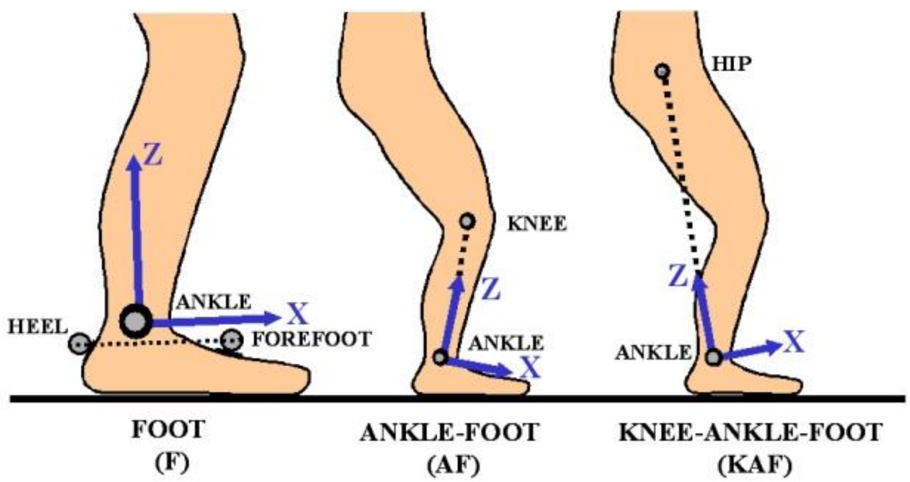
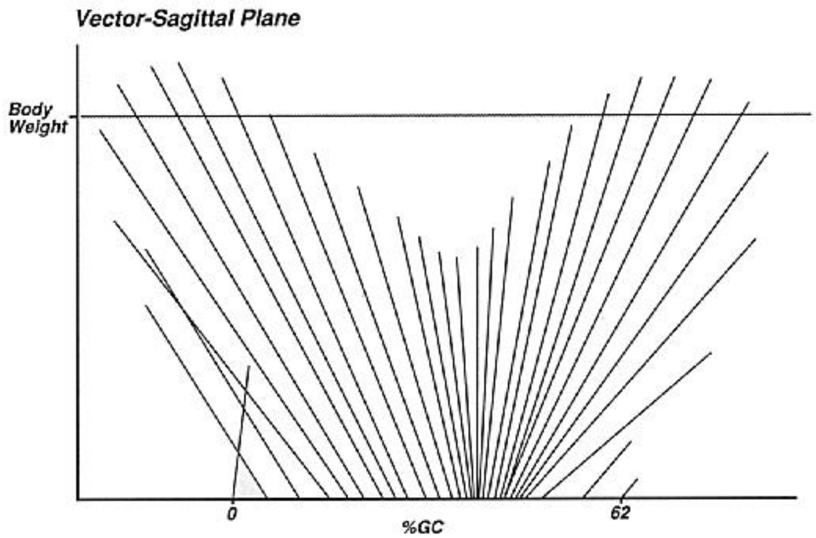


Figure A.2: Marker placements for 3 marker-based coordinate systems to determine roll-over shape (Hansen, 2000).



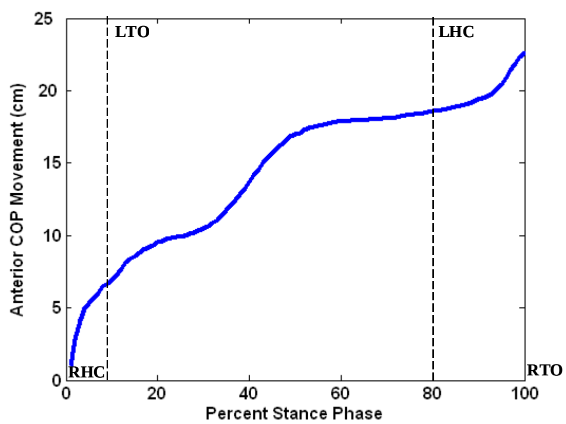


Figure A.3: Ground reaction force vectors over the gait cycle (top) (Perry 1992) and corresponding movement of the center of pressure (COP) in the sagittal plane (direction of forward progression) over the stance phase of gait (one step), or 0 to 62% of the gait cycle (bottom). The COP sagittal plane view shows that at the beginning of the step, the COP, and thus the foot, rolls quickly forward about 17cm and then pauses around 50% of the step while the body mass moves over the foot (similar to an inverted pendulum). The COP moves forward again just before toe off and the end of the step. The GRF vectors, which are taken at equal time intervals, also show the pause of the GRF around 31% of the GC (50% of stance phase).

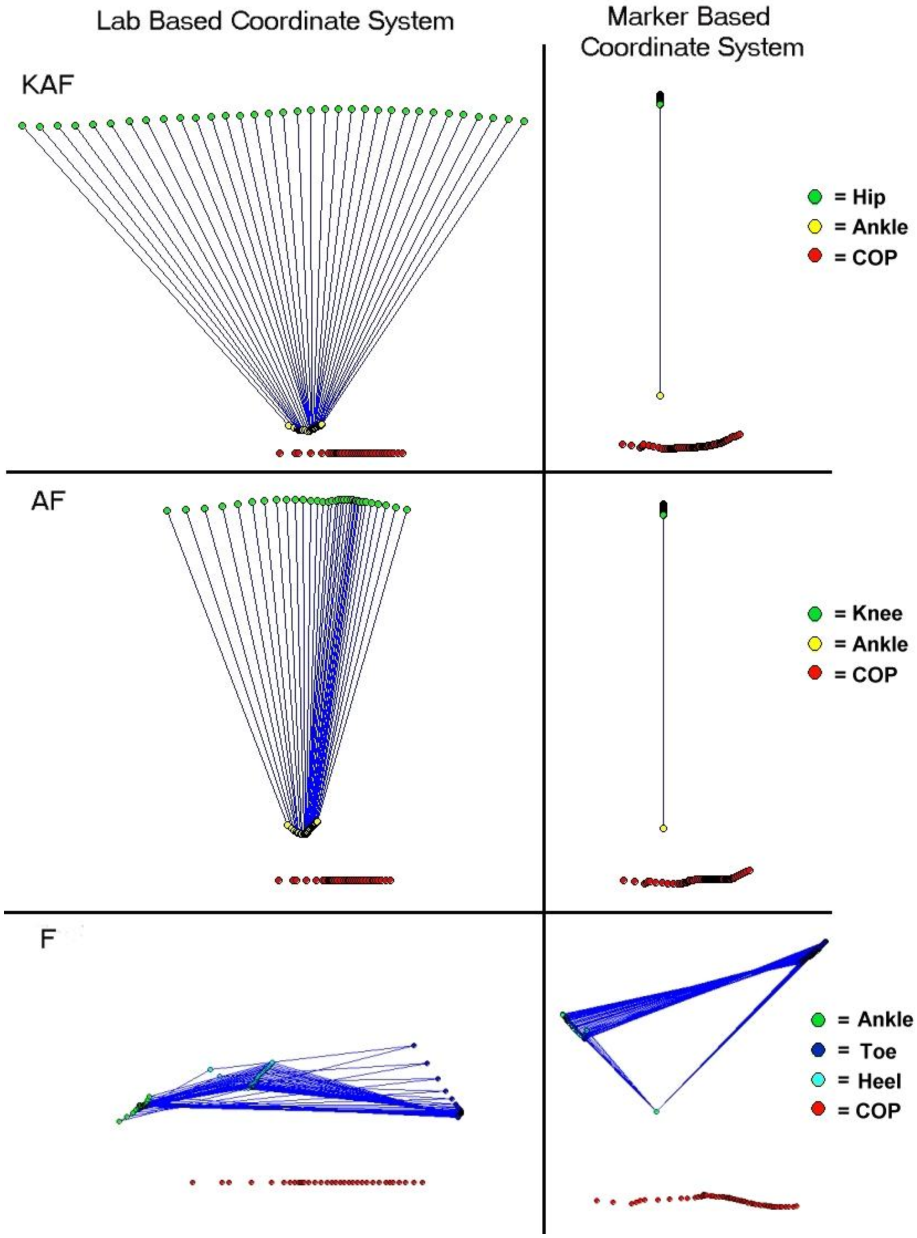


Figure A.4: Transformations of COP into marker-based coordinate systems. Left side shows COP and markers used for transformation in a world-based coordinate system for normal walking from heel contact (HC) to opposite HC. Right side plots show transformations into marker-based coordinate system. For KAF, COP is transformed into an Ankle-Hip coordinate system. For AF, COP is transformed into a shank-based (Ankle-Knee) coordinate system. The foot roll-over shape uses foot-based coordinate system to transform COP. A more detailed view of the AF system is shown in Figure A.5

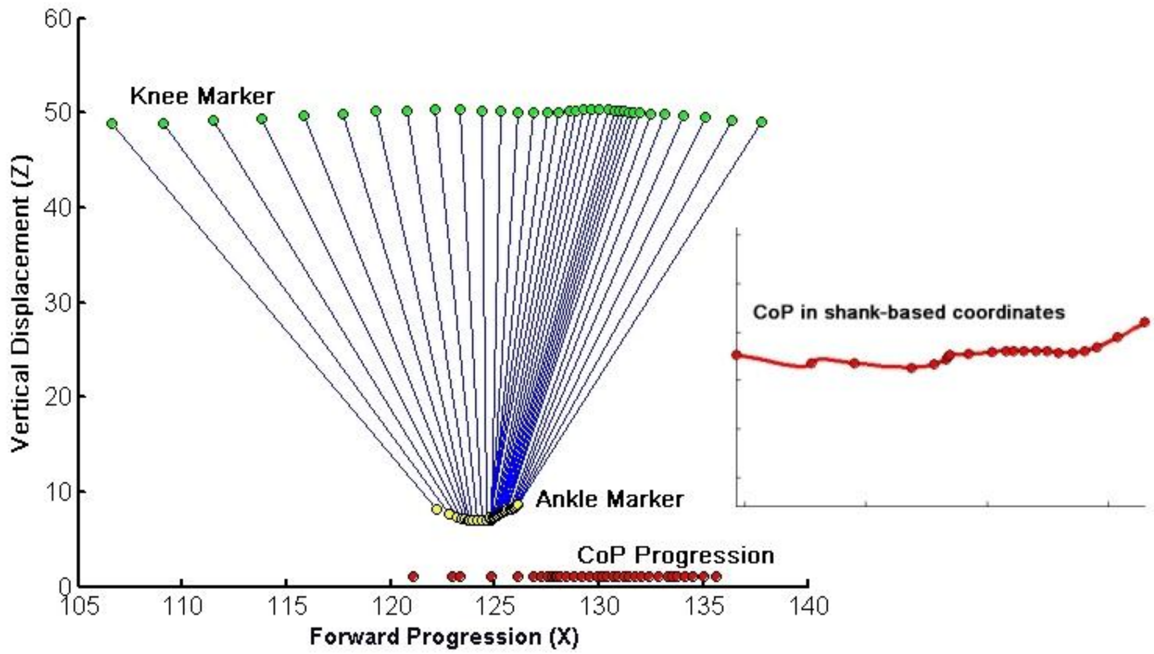


Figure A.5: Detailed view of markers and COP used for AF shapes in a world-based coordinate system. Inset shows the final roll-over foot shape in a shank-based coordinate system during self-selected walking.

In the world-based coordinate system, the COP is the location of the base of the ground reaction force (GRF) vector produced when a body segment is in contact with the floor (usually the foot) (Figure A.4). The GRF is a mean force vector that represents the loading of the body on the ground. The COP can represent the rolling shape that the lower limb joints conform to when transformed into a marker-based coordinate system. The marker-based coordinate system is comprised of specific markers used for capturing motion of the joints and segments of the body. Markers used for the coordinate system stay relatively rigid. For the AF roll-over shape, the markers used are at the ankle and knee; and for the KAF roll-over shapes, the markers used are at the ankle and hip (Figure A.2).

In the marker-based coordinate system, the axes created by these markers (X and Z axes for the sagittal plane) are kept stationary while the COP position is transformed relative to these markers to create the roll-over shape. In the AF marker-based coordinate system, the Z-component is the vector pointing upwards from the ankle marker through the knee marker. The X-component is the vector perpendicular to the Z-component in the sagittal plane and points “forward” (towards the toes). Figure A.5 displays an example of this transformation.

For the foot shape, a small initial rollover curved shape can be seen, comparable to the heel-rocker suggested by Perry (1992). The AF shape shows a concave up curve, similar to a rocker shape, throughout the stance phase. This may be attributed to adaptation of the ankle joints to create the rollover foot shape.

Utilizing the rollover foot shape, a measure of the length of the foot that actually bears weight (the area of the foot below which the COP lies) while walking can be measured. This is named the “effective foot length” (EFL) and is measured as the length of the roll-over foot shape, or the distance from the heel to the anterior end of the shape (Hansen, Sam et al. 2004). The effective foot length ratio (EFLR) is described as the ratio of the EFL over the total foot length (FL):

$$EFLR = \frac{EFL}{FL}$$

where FL is the distance from the heel to the toes. During normal walking, Hansen et al. (2004) calculated the EFLR of the physiologic foot to be approximately 0.83 using data from 24 able-bodied persons walking at speeds between 1.20 and 1.60 meters/sec. The EFL was calculated by adding the sagittal distance between the heel and ankle ($l_{posterior\_foot}$ = 0.26 × FL) to the distance between the ankle and anterior end of the foot shape ($l_{anterior\_footshape}$ = 0.52 × FL). The heel to ankle measurement was determined using anthropometric data from Dreyfuss (1967) since it could not be measured from the foot shape (initial COP data at heel contact is not accurate and thus the posterior end of the rollover shape may be inaccurate (Hansen, Sam et al. 2004)). The total FL was also determined from anthropometric data using the subject's height, where FL = 0.152 *H* (Dreyfuss 1967; Winter 1990).