THE EFFECTS OF WALKING STEP RATE ON
GROUND REACTION FORCES AND MUSCLE ACTIVATION

by

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A thesis submitted to the Faculty of the University of Delaware in partial fulfillment of the requirements for the degree of Master of Science in Biomechanics and Movement Sciences

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My professors, both through my undergraduate and graduate education, who have helped me find the right academic path.

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ABSTRACT

It is well known that metabolic cost is minimized when walking at a preferred step rate. However, the relationship between step rate and other variables such as ground reaction forces and muscle activity is not well established. If altered step rates lead to increased forces, it could help to explain why certain populations (such as unilateral lower-extremity amputee subjects) experience increased joint loading on the intact limb. Higher joint forces are especially harmful to amputee subjects, who often develop knee osteoarthritis in their intact limb. The purpose of this study was to examine ground reaction forces and muscle activity associated with walking at step rates above and below preferred. In Part 1 of the study, 20 subjects walked under 7 step rate conditions at a constant speed of 1.3 m/s. The step rate conditions were preferred and -30%, -20%, -10%, +10%, +20%, and +30% steps/min from the preferred. In Part 2, subjects walked at the same step rates from Part 1, but step length was held constant by adjusting the treadmill speed for each condition. Motion capture, ground reaction forces, and EMG on 5 muscle groups (tibialis anterior, vastus lateralis, biceps femoris, medial head of gastrocnemius, and soleus) were recorded. A series of one-way repeated measures ANOVAs found significant (p<0.05) differences in all variables for both Parts. A U-shaped curve with the minimum at the preferred condition was found for the vertical ground reaction force, loading rate, and EMG data in Part 1. In Part 2, the same variables increased linearly with increasing step rate.
These results show that at higher step rates (both with constant walking speed and constant step length) the ground reaction forces and muscle activity increase.

Clinically, this is important because of the implications toward increased limb loading seen in amputee subjects and may help to explain why there is a higher rate of knee osteoarthritis in their intact limb.
Chapter 1
INTRODUCTION

It is widely believed that certain walking parameters minimize the metabolic cost of walking; when people deviate from their self-selected step length (SL) and step rate (SR), energy expenditure increases (Holt et al., 1991). There may be factors other than metabolic cost that cause people to choose certain walking patterns. Perhaps ground reaction forces (GRF) are minimized at the self-selected step length and step rate during walking. Diverging from the self-selected step length and preferred step rate (PSR) has the potential to cause increases in the ground reaction forces during walking (Martin and Marsh, 1992, Russell et al., 2010, Soames and Richardson, 1985). A higher than normal GRF is associated with some of the abnormalities observed in pathological gait and may be a precursor to orthopedic injury.

Several studies have looked at how GRF changes with different step rates (Martin and Marsh, 1992, Russell et al., 2010, Soames and Richardson, 1985). Martin and Marsh (1992) found significant differences in contact time, braking and propulsive peak forces, and impulse at step rates lower than the PSR; however there were no significant changes in vertical GRF, most likely due to the minor manipulations of step rate (around 10% from PSR). Russell et al (2010) found small reductions in the vertical ground reaction forces (GRFv) with increasing SR. Step
length was decreased by 15% (and therefore step rate increased) in obese and healthy-weight subjects, resulting in decreased GRF. The manipulations may have been too minor in order to see significant changes in the vertical GRF impact and knee adduction moments, as the researchers suggested. By using larger manipulations of step rate, significant changes in GRFv may be found; Soames and Richardson (1985) used a broader range of SR and found that both SL and SR have significant effects on the GRF. While GRFv increased in parallel with increasing step rate, the peak GRF in all directions were only significant between the slowest and fastest step rates, 84 steps/min and 124 steps/min. The results from the Soames and Richardson (1985) study further support the idea that the manipulation in step rate needs to be pronounced in order to see significant changes.

Higher GRFs lead to higher loading at the knee which may cause overuse injury such as knee osteoarthritis (OA) (Amin et al., 2004, Zhao et al., 2007). Osteoarthritis is the most common form of arthritis, affecting 13.9% of adults over the age of 25 years and 33.6% above 65 years (Centers for Disease Control and Prevention, Hootman et al., 2006). Knee OA affects around 16% of adults above the age of 45 years, and about 80% of those with knee OA have movement limitation. There are several factors that researchers suggest could lead up to or exacerbate the development of knee OA. One of which is an increased internal knee abduction moment (Amin et al., 2004, Zhao et al., 2007). The abduction moment is substantial during walking and accounts for a large portion of the loading at the knee, which carries 60-70% of the weight-bearing forces during walking (Andriacchi, 2004). With
a greater GRF, the abduction moment increases, and consequently the medial loading of the knee also increases.

People who have suffered a unilateral transtibial amputation are at an even greater risk of developing knee OA in their intact leg, and have a greater than 65% reported incidence of OA, notably higher than the healthy population (Melzer et al., 2001). An amputee subject’s intact limb experiences greater forces and develops increased bone mineral density (BMD) on the proximal tibia (Burger et al., 1996, Burke et al., 1978, Royer and Koenig, 2005), both of which are potential precursors of knee OA. The imbalance of forces is a result of the asymmetrical gait patterns that amputee subjects exhibit (Dingwell et al., 1996, Sanderson and Martin, 1997). The increased stress resulting from the higher GRF acting on the intact limb of amputee subjects contributes to the degradation of cartilage, tendons, and ligaments, (Hamill et al., 1999). With a higher GRF, joint moment, or muscle force, there is an increase in bone loading at the knee joint, which raises the risk of developing OA at that joint.

In this study, walking SR is defined as the reciprocal of the time it takes from the contralateral heel strike to ipsilateral heel strike. For example, the SR of the right limb is the reciprocal of the time from left heel strike to right heel strike. By this definition, it follows that SR is mainly influenced by swing time; a faster swing phase would result in a higher SR. Another approach to calculating SR is based on its relationship with walking velocity and SL:

\[ SR = \frac{SL}{Velocity} \]
Both of these methods would result in the calculation of an asymmetrical between limb step rate for a unilateral amputee subject, because of the commonly observed between limb spatiotemporal asymmetries. A quicker swing phase and longer step length for the intact limb results in a higher step rate for the intact limb compared to the prosthetic side. In reality, both limbs take the same number of steps per unit time when walking in a straight line. These standards for calculating SR will be used in this study.

Unilateral, lower-extremity amputee subjects favor their intact limb, resulting in a greater stance time on that side; this leads to kinetic and spatiotemporal asymmetries. The prosthetic limb has a smaller stance time and longer swing time (Sanderson and Martin, 1997, Isakov et al. 1996), most likely due to unloading their prosthetic side due to issues such as pain, instability, weakness, or loss of sensory feedback. Stepping onto the more stable, intact limb more quickly results in a higher step rate for the intact limb. The intact limb has an increased stance time and decreased swing time, so the prosthetic side swings through slowly and has a smaller step rate. It is possible the faster SR of the intact limb contributes to higher loading and risk of developing knee OA in that limb.

Persons with a unilateral amputation experience a greater peak GRF in their intact limb as compared to their prosthetic limb (Sanderson and Martin, 1997, Beyaert et al 2008, Nolan et al., 2003). Beyaert et al. (2008) found the GRF of the intact limb to be 14% higher compared to prosthetic limbs, and 9% higher than the control limbs. Nolan et al. (2003) had similar findings, and also found that the differences in GRF
values increased with increasing walking speed. Other studies support these findings, though with less pronounced increases in the GRF. Sanderson and Martin, 1997 found a 1% increase in the intact limb vertical ground reaction force (GRFv) compared to the prosthetic limb, while the intact limb’s GRFv were 3% higher than healthy control subjects.

Amputee and stroke patients, or anybody that has a condition that affects the lower limbs unilaterally (such as an injury or wearing a brace), tend to favor the unaffected limb and often show higher forces in this limb. Because of the problems that develop with the healthy limb in pathological gait, it is important to understand the underlying mechanisms of the increased forces that the unaffected limb experiences. SR may play a key role in exacerbating lower extremity loading.

While it is established that walking at step rates outside the PSR increases metabolic cost, it is unclear how GRF data are related to changes in SR. If a faster step rate results in increases in forces, this would help to explain why the intact limb of unilateral lower-extremity amputee subjects experiences increased loading. Increases in loading at the knee are especially detrimental to amputee subjects because they will often develop knee osteoarthritis (OA) in their intact limb (Melzer et al., 2001). The purpose of this project is to examine GRF and muscle activity associated with walking at step rates above and below preferred. It was hypothesized that ground reaction forces and muscle activation would be minimized 1) at a preferred step rate when walking speed is constant and 2) at the lowest step rate when step length is held constant. Forces and muscle activity will be greater when walking at step rates above
and below preferred (while walking speed is constant); however, these variables will have a positive linear relationship with step rate when step length is held constant.
Chapter 2

METHODS

2.1 Subjects

20 able-bodied individuals, 8 males and 12 females, were recruited for this study. Subjects were free of lower extremity injuries and were comfortable walking on a treadmill. The average age of the subjects was 21.8±3.4 years, height was 1.7±0.1 m, and mass was 66.4±12.3 kg. All subjects signed informed consent documents approved by the University of Delaware HSRB.

2.2 Walking Condition Descriptions

This study involved two parts in which SR was increased and decreased from the preferred step rate. In Part 1, walking speed was held constant as step length increased with smaller step rates and decreased with larger step rates. In Part 2 step length was held constant as walking speed was reduced for smaller step rates and increased for larger step rates.

Subjects walked under 7 SR conditions where walking speed was constant, and 7 SR conditions where SL was constant for a total of 14 conditions. Subjects walked on an instrumented treadmill (Bertec Corp., Columbus, OH) in the Mechanical
Engineering Treadmill Lab at the University of Delaware. This treadmill measures GRF in three dimensions. In Part 1, the subject walked with a self-selected SR at a speed of 1.3 m/s; this is the nominal condition from which other conditions were created. During this condition, the PSR was calculated by using a stopwatch to measure the time it took for the subject to complete 50 steps. The PSR was calculated with the equation:

$$PSR(\text{steps/min}) = \frac{50\text{steps}}{\text{time (sec)}} \cdot 60\text{sec/min}$$

The other six conditions consisted of the subject walking with different step rates at the constant speed of 1.3 m/s. These conditions were -10%, -20%, -30%, +10%, +20%, and +30% steps/min from the PSR.

In Part 2, the subject walked at the same step rates from Part 1, but the step length was held constant by changing the treadmill speed for each condition. To achieve this, first the step length was calculated from the PSR in Part 1 with the equation:

$$\text{Step length (m)} = \frac{1.3m/s}{\text{step rate (steps/min)}} \cdot 60\text{sec/min}$$

then the appropriate treadmill speed was found using:

$$\text{Treadmill speed (m/s)} = \text{step length (m)} \cdot \text{step rate (steps/min)} \cdot \frac{60\text{sec/min}}{60\text{sec/min}}$$
2.3 Data Capture

To record three-dimensional motion of the lower extremities, 12.7 mm diameter retroreflective markers were placed at the following locations on both limbs: 1st and 5th metatarsal heads, heel, medial and lateral ankle, shank, medial and lateral knee, thigh, greater trochanter, ASIS, and sacrum. An 8 camera motion analysis system (Motion Analysis, Santa Rosa, CA) captured the locations of the markers at 60 Hz.

The force plates embedded in the treadmill recorded the GRF at a rate of 1080 Hz. An electromyography (EMG) system (Noraxon, Scottsdale, AZ) was used to collect surface EMG signals at 1080 Hz from five different muscle groups: tibialis anterior (TA), vastus lateralis (VL), biceps femoris (BF), medial head of gastrocnemius (MG), and the soleus (SO). A reference electrode was placed on the tibial plateau. EMG collection was done unilaterally, on the right leg. Electrodes were affixed in parallel with the muscle fibers over the distal half of the muscle belly. In order to reduce impedance, the skin of each electrode site was shaved, abraded, and cleaned with isopropyl alcohol. Subjects were given a short acclimation period (2 minutes) for each condition and data collection consisted of two trials of 30 seconds under each condition. The motion, GRF, and EMG data capture were synchronized. In the case where a subject had difficulty maintaining a designated step rate, the trial collection time was reduced to 15 seconds. Subjects maintained the different step rates
by matching each heel strike to the beat of a metronome set to the prescribed SR. These seven conditions within each Part were randomized for each subject.

### 2.4 Data Processing

Visual 3D software (C-Motion, Inc., Germantown, MD) was used to filter the force data, identify heel strike and toe off events, and calculate step length data. The GRF data were filtered with a lowpass Butterworth filter with a cutoff freq of 40 Hz. The filtered data were then run through a customized LabView (National Instruments, Austin, TX) program that normalized the GRF data to each subject’s body weight, identified the peak GRFv and GRFap, and then averaged the peak values within each condition for each subject. For clarity, the peak GRFap represents the peak propulsive force (i.e., the reaction force in the anterior direction). LabView was also used to calculate the GRFv loading rate (LR), by taking the slope of the GRFv between 20% and 80% of the first peak (Butler et al, 2006). A depiction of how the GRFv LR was calculated is shown below in Figure 1.
Figure 1: The GRFv LR, depicted by the bold line, was calculated by taking the slope between the interval from 20% to 80% of the first peak of the GRFv.

The EMG data were processed with a separate LabView program. This program used a typical processing algorithm of removing the DC bias, bandpass filtering the signal (with a low cutoff frequency of 20 Hz and high cutoff frequency of 500 Hz), full wave rectify, and creating a linear envelope (with a low pass filter of 10 Hz). Additionally, data were normalized in the following manner: an EMG ensemble average for each muscle was calculated using approximately 25 stride cycles from the PSR condition. The peak EMG value from this ensemble average was the normalization value for each muscle (Yang and Winter, 1984). EMG data for each condition were normalized to this value. EMG data were visually inspected for gross errors (e.g., excessive noise, improper time normalization, etc). For each stride cycle, the peak value for each muscle group for both stance (EMGst) and swing phases
(EMGsw) was identified, and then averaged within each condition. The result is an average EMGst and EMGsw for each subject, each muscle, and each condition.

Kinematic data were tracked and smoothed (Butterworth filter with frequency of 6 Hz) in Cortex (Version 2.5, Motion Analysis Corp., Santa Rosa, CA). Visual 3D software provided the knee flexion angle of the right leg, which was then run through a customized LabView program to obtain the knee angle excursion. Knee angle excursion was found by calculating the difference in knee flexion angle from the point of heel strike to the point of peak GRFv.

2.5 Statistical Analysis

Analysis of each subject was conducted using data collected from the right leg. A repeated measures design was used for statistical analysis since the design of this study uses comparisons within subjects. A series of one-way repeated measures ANOVAs (with 7 conditions) was calculated within each Part of this study. Within each dependent variable, preplanned contrasts compared pairs of adjacent conditions and compared the PSR condition to all other conditions. The variables of interest are the peak GRFv, GRFap, GRFv LR, knee flexion angle excursion, and EMGst and EMGsw for five muscles. A two-factor ANOVA, SR condition (-30%, -20%, -10%, preferred, +10%, +20%, +30%) by SL type (targeted, actual) tested for differences between the actual and targeted SL for Part 2. An α level of 0.05 was used for all analyses. SPSS (Version 20.0, IBM Corp., Armonk, NY) was used for all of the statistical tests.
3.1 Part 1 – Constant Walking Speed

Paired t-tests showed a significant difference between the subjects preferred walking speed (1.22±0.1 m/s) and the prescribed walking speed of 1.3 m/s (p<0.01). The SR and SL data for Part 1 are shown in Table 1. The differences between targeted and actual SR were minor. Since walking velocity was held constant, SL varied and decreased as SR increased.

Table 1: Average and standard deviation values for SR and SL data from Part 1. The targeted and actual values for SR are shown, as well as differences between them.

<table>
<thead>
<tr>
<th>SR Condition</th>
<th>Targeted SR (steps/min)</th>
<th>Actual SR (steps/min)</th>
<th>Difference in SR (steps/min)</th>
<th>SL (m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>-30%</td>
<td>79.8 (3.8)</td>
<td>82.3 (7.8)</td>
<td>3.2 (4.3)</td>
<td>0.95 (0.07)</td>
</tr>
<tr>
<td>-20%</td>
<td>91.2 (4.3)</td>
<td>93.6 (7.5)</td>
<td>2.9 (3.8)</td>
<td>0.84 (0.06)</td>
</tr>
<tr>
<td>-10%</td>
<td>102.5 (4.6)</td>
<td>102.8 (5.7)</td>
<td>1.2 (0.9)</td>
<td>0.76 (0.04)</td>
</tr>
<tr>
<td>Preferred</td>
<td>114.0 (5.2)</td>
<td>113.9 (5.5)</td>
<td>0.8 (0.6)</td>
<td>0.69 (0.03)</td>
</tr>
<tr>
<td>+10%</td>
<td>125.3 (5.7)</td>
<td>125.3 (5.8)</td>
<td>1.4 (1.8)</td>
<td>0.63 (0.03)</td>
</tr>
<tr>
<td>+20%</td>
<td>136.7 (6.3)</td>
<td>134.7 (6.5)</td>
<td>2.9 (4.9)</td>
<td>0.58 (0.02)</td>
</tr>
<tr>
<td>+30%</td>
<td>148.1 (6.8)</td>
<td>145.7 (7.0)</td>
<td>3.3 (4.8)</td>
<td>0.54 (0.02)</td>
</tr>
</tbody>
</table>

There were significant (p<0.01) differences in GRFv among the SR conditions. Specifically, the GRFv values for the -30%, -20%, and the +30% SR conditions were
all significantly greater than the PSR (p<0.01). The GRFv followed a U-shaped pattern with the minimum, 1.17±0.05 BW, occurring at the PSR (Figure 2). The highest GRFv value of 1.29±0.11 BW (which is 10% greater than PSR) occurred at -30% of the PSR. This value is slightly higher than the GRFv at the +30% condition, which was 1.22±0.07 BW.

The GRFap values also showed significant (p<0.01) differences. All of the values for GRFap were significantly different from the PSR condition (p<0.01). The GRFap data followed a negative linear pattern, as seen in Figure 3. The highest value for GRFap, 0.31±0.03 BW, occurred at -30%, and the lowest value, 0.18±0.02 BW, occurred at the +30% SR condition.

There were significant (p<0.01) differences in the GRFv LR data. All of the values for GRFv LR were significantly different from the PSR (p<0.05). The GRFv LR values followed a U-shaped pattern, similar to GRFv (Figure 4). However, the lowest value for GRFv LR, 8.26±1.27 BW/sec, occurred at -10%, instead of the PSR. The PSR GRFv LR had a slightly higher value of 8.59±1.19 BW/sec. The highest value was 12.07±1.78 BW/sec (which was 40% greater than PSR), which occurred at the +30% SR condition.

The EMGst values had significant (p<0.05) differences among the different SR conditions (Figure 5). Significant differences from the PSR condition were found in TA for -30%, -20%, and +20%; in VL and BF for -30% and -20%; in MG for -20%; and in SO for -30%, -20%, and -10% SR conditions (p<0.05). EMGst values were
minimized at PSR for the VL, and MG muscle groups, and were close to minimization for the remaining groups (Figure 5). All muscle groups also showed a U-shaped trend.

There were significant (p<0.05) differences among the EMGsw values. EMGsw values were only minimized at the PSR for the SO muscle group, however the TA, VL, and SO groups all showed a U-shaped pattern (Figure 6). The BF and MG muscle groups had a more linear pattern, with values increasing with increasing SR. Significant differences (p<0.05) from the PSR were found in the TA for the -30% and -20% conditions, VL and BF muscle groups for the +10%, +20%, and +30% conditions, and in the MG muscle group for +20% and +30% conditions (p<0.05).

The knee flexion angle excursion data had significant (p<0.01) differences among the different SR conditions; all of the different SR conditions were significantly (p<0.05) different from the PSR (Figure 7). Knee flexion angle excursion decreased almost linearly with increasing SR. Its maximum value of 21.6±4.7º occurred at the -30% SR condition and decreased down to a minimum of 9.2±3.5º at the +30% SR condition.
Figure 2: The peak GRFv with standard deviation bars for constant walking speed (Part 1). There was a trend for peak GRFv to be minimized at the PSR. (a=sig diff (p<0.01) from previous condition. b=sig diff (p<0.01) from PSR)

Figure 3: The peak GRFap with standard deviation bars for constant walking speed (Part 1). The GRFap values showed a negative linear trend with walking speed. (a=sig diff (p<0.01) from previous condition. b=sig diff (p<0.01) from PSR)
Figure 4: The GRFv LR with standard deviation error bars for constant walking speed (Part 1). The GRFv LR followed a U-shaped trend, with the minimum occurring at -10% condition. (a=sig diff (p<0.01) from previous condition. b=sig diff (p<0.01) from PSR)

Figure 5: EMGst values for all 5 muscle groups for Part 1. Within each muscle group, the 7 walking conditions are shown, starting from -30% (on the left) and going in
increasing value up to +30% (farthest right). Error bars are standard deviation. (a=sig diff (p<0.01) from previous condition. b=sig diff (p<0.01) from PSR)

Figure 6: EMGsw values for all 5 muscle groups for Part 1. Within each muscle group, the 7 walking conditions are shown, starting from -30% (on the left) and going in increasing value up to +30% (farthest right). Error bars are standard deviation. (a=sig diff (p<0.01) from previous condition. b=sig diff (p<0.01) from PSR)
Figure 7: Knee flexion angle excursion (from heel contact to peak GRFv) with standard deviation error bars for constant walking speed (Part 1). With increasing SR, the knee flexion angle excursion decreased. (a=sign diff (p<0.01) from previous condition. b=sign diff (p<0.01) from PSR)

3.2 Part 2 – Constant Step Length

The SR and SL data for Part 2 are shown in Table 2. The targeted SR values are omitted because they are the same values from Part 1. All of the differences between targeted and actual values for SR and SL are minor. A two-factor ANOVA, SR condition (-30%, -20%, -10%, preferred, +10%, +20%, +30%) by SL type (targeted, actual), resulted in a significant (p<0.01) interaction between SR condition and SL type. Post hoc t-tests resulted in significant (p<0.05) differences between the targeted and actual SL for the following conditions: -30%, +10%, +20%, and +30%. Of those, the largest difference was found between the targeted and actual SL for the +30% condition; however the difference was on average only 2.2 cm. Even with these differences, all of the subjects were able to maintain a SL within centimeters of the targeted value.
Table 2: Average and standard deviation values for SR and SL data from Part 2. The targeted SR are the same values from Table 2.

<table>
<thead>
<tr>
<th>SR Condition</th>
<th>Actual SR (steps/min)</th>
<th>Difference in SR (steps/min)</th>
<th>Targeted SL (m)</th>
<th>Actual SL (m)</th>
<th>Difference in SL (m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>-30%</td>
<td>81.0 (4.2)</td>
<td>2.2 (2.7)</td>
<td>0.69 (0.03)</td>
<td>0.68 (0.03)</td>
<td>0.01 (0.01)</td>
</tr>
<tr>
<td>-20%</td>
<td>91.0 (4.7)</td>
<td>0.8 (0.5)</td>
<td>0.69 (0.03)</td>
<td>0.69 (0.04)</td>
<td>0.01 (0.02)</td>
</tr>
<tr>
<td>-10%</td>
<td>102.2 (5.0)</td>
<td>0.8 (0.6)</td>
<td>0.69 (0.03)</td>
<td>0.69 (0.03)</td>
<td>0.01 (0.00)</td>
</tr>
<tr>
<td>Preferred</td>
<td>114.0 (5.6)</td>
<td>0.6 (0.4)</td>
<td>0.69 (0.03)</td>
<td>0.69 (0.03)</td>
<td>0.00 (0.00)</td>
</tr>
<tr>
<td>+10%</td>
<td>124.8 (5.8)</td>
<td>0.9 (0.9)</td>
<td>0.69 (0.03)</td>
<td>0.69 (0.03)</td>
<td>0.01 (0.00)</td>
</tr>
<tr>
<td>+20%</td>
<td>134.6 (6.4)</td>
<td>2.7 (3.3)</td>
<td>0.69 (0.03)</td>
<td>0.70 (0.03)</td>
<td>0.01 (0.02)</td>
</tr>
<tr>
<td>+30%</td>
<td>145.0 (5.4)</td>
<td>3.7 (4.6)</td>
<td>0.69 (0.03)</td>
<td>0.71 (0.02)</td>
<td>0.02 (0.02)</td>
</tr>
</tbody>
</table>

There were significant (p<0.01) differences for the GRFv data among the different SR conditions; all of the conditions were significantly different from the PSR (p<0.01). For constant step length, the GRFv showed a positive linear trend with increasing SR and walking speed (Figure 8). The values ranged from the minimum value of 1.06±0.04 BW at the -30% condition up to 1.32±0.07 BW at the +30% condition.

The GRFap followed a similar trend to the GRFv values, but with a smaller range (Figure 9). There were significant (p<0.01) differences among the GRFap values, and all of the values were significantly different from the PSR (p<0.01). The values ranged from 0.19±0.02 BW at the -30% up to 0.25±0.03 BW at the +30% condition.

There was a significant difference (p<0.01) in GRFv LR among the SR conditions; all of the GRFv LR values were also significantly different from the PSR (p<0.01). Again, the GRFv LR followed the same linear pattern as GRFv and GRFap (Figure 10). There was a positive linear relationship between SR and GRFv LR, with a
minimum of 4.68±1.19 BW/sec at -30% and a maximum of 14.75±1.62 BW/sec at +30%.

Significant (p<0.05) differences for EMGst data were found (Figure 11). Specifically, significant differences from the PSR condition were found in TA for +30%, in VL for -30%, -20%, +10%, +20%, and +30%; in MG for +10%, +20%, and +30%; and in SO for -30%, -20%, +10%, and +30% SR conditions (p<0.05). The stance EMG data from Part 2 are shown in Figure 11. EMGst values followed a trend of increasing with increasing SR.

The EMGsw data also had significant (p<0.05) differences among the different SR conditions; significant differences (p<0.05) from the PSR condition were found in TA for +10%, +20%, and +30%; in VL for -30%, -20%, +10%, +20%, and +30%; in BF for all of the conditions; in MG for -30%, -20%, -10%, and +30%; and in SO for -20%, -10%, and +30% SR conditions (Figure 12). EMGsw data also showed a positive linear trend with increasing SR.

The knee flexion angle excursion data had significant (p<0.01) differences among the different SR conditions; all of the different SR conditions except for +10% and +20% were significantly (p<0.05) smaller than the PSR (Figure 13). Knee flexion angle excursion increased almost linearly from -30% up to the PSR and then appeared to decrease from the PSR to +30%. The minimum excursion occurred at -30% with a value of 9.6±6.1°, while the maximum value of 16.9±5.0° occurred at the PSR.
Figure 8: The average peak GRFv with standard deviation bars for constant step length (Part 2). With a changing treadmill speed, the GRFv values showed a positive linear trend with SR and walking speed. (a=sig diff (p<0.01) from previous condition. b=sig diff (p<0.01) from PSR)

Figure 9: The average GRFap with standard deviation bars for constant step length (Part 2). Similar to the GRFv, the GRFap showed a positive linear trend with SR and
walking speed. (a=sign diff (p<0.01) from previous condition. b=sign diff (p<0.01) from PSR)

Figure 10: The average GRFv LR with standard deviation bars for constant step length (Part 2). Similar to the GRFv and GRFap, the LR showed a positive linear trend with SR and walking speed. (a=sign diff (p<0.01) from previous condition. b=sign diff (p<0.01) from PSR)
Figure 11: EMGst values for all 5 muscle groups for Part 2. Within each muscle group, the 7 walking conditions are shown, starting from -30% (on the left) and going in increasing value up to +30% (furthest right). Error bars are standard deviation. (a=sig diff (p<0.01) from previous condition. b=sig diff (p<0.01) from PSR)
Figure 12: EMGsw values for all 5 muscle groups for Part 2. Within each muscle group, the 7 walking conditions are shown, starting from -30% (on the left) and going in increasing value up to +30% (furthest right). Error bars are standard deviation. (a=sig diff (p<0.01) from previous condition. b=sig diff (p<0.01) from PSR)

Figure 13: Knee flexion angle excursion (from heel contact to peak GRFv) with standard deviation error bars for constant SL (Part 2). From the -30% SR condition to the PSR, knee flexion angle excursion increased with increasing SR and walking speed. From the PSR to the +30% SR condition, knee flexion angle excursion slightly decreased. (a=sig diff (p<0.01) from previous condition. b=sig diff (p<0.01) from PSR)
Chapter 4

DISCUSSION

The goal of this study was to observe how GRF and EMG data are affected by walking at SR above and below the PSR. There was a trend for GRFv, GRFv LR, and EMG data to be minimized around the PSR when speed was held constant (Part 1). With SL held constant (Part 2), GRF and EMG values all increased with increasing SR and walking speed.

As the results show in Table 7, subjects were successful in matching their prescribed SR. In Part 1, there was an average of around 2% difference between prescribed and actual SR. Unsurprisingly, the largest differences were seen in the SR of the -30% condition, which was a 4.5% difference from preferred. Subjects were also successful in matching both their prescribed SR and SL in Part 2, as the results indicate in Table 8. There was an average difference in the targeted and actual SR of only 1.5%. For the SL data, there was an average difference of 3% between the targeted and actual values. These results show that subjects are able to match their heel strikes with the metronome and the methods for controlling SR and SL are accurate.

The resulting GRFv, GRFv LR, and EMG data follow the same trends as the metabolic cost data from the Holt study (Holt et al, 1991). With the constant walking speed (Part 1), Holt also found a parabolic relationship for oxygen consumption and
SR, with the minimum occurring at the PSR and slightly higher values of oxygen consumption at the lower SR (compared to those above the PSR). While the GRFv value at -30% was not significantly different than the +30%, it did have slightly higher values, following the published data. The GRFv LR also followed the expected trend, with the exception being that the minimum occurred at -10% instead of at the PSR. The two values were close, however, and had less than a 4% difference. It has been shown that a knee flexion angle >15° can contribute to mechanical overloads (Harato et al 2008), which means higher muscle activity in VL and BF leads to higher LR. Since GRFv LR and VL and BF activity all followed the U-shaped trend, there is evidence in this study to support those results. The GRFap data followed a linear trend, with higher values occurring at lower SR (and therefore longer SL) and lower values at higher SR (and shorter SL). This trend follows the findings of Martin and Marsh (1992), which found that peak GRFap increased systematically with decreasing SR.

EMG data also followed the U-shaped trend with walking SR. To increase SR, MG and SO activity would increase during the stance phase and TA, BF and VL activity would increase during the swing phase. While MG and SO only showed minor increases during the stance phase for increasing SR, VL and BF showed the expected trends. During stance, EMG activity increased more for SR below the PSR, indicating that lower SR have higher musculature demand; indeed, subjects reported having more difficulty in maintaining the lower SR compared to the higher ones. The EMG data
may help to explain the increases in metabolic cost when walking at SR above and below the PSR in the Holt study (Holt et al., 1991).

At a constant walking speed, the knee flexion angle excursion decreased with increasing SR. Kinematically this is unsurprising as it follows the trend for SL. With a longer SL/lower SR, the leg is going through a greater range of motion, which would increase the knee flexion angle excursion. At a higher SR, SL is shortened, and the knee angle excursion decreases.

For Part 2 SL was held constant; Holt found a positive linear trend between oxygen consumption and SR (as well as walking velocity). All of the GRF data (GRFv, GRFap, and GRFv LR) showed increasing values with increasing SR. This is in accordance with Soames and Richardson (1985) who found that the GRFv and GRFap increased with increasing SR. Other studies have found that higher walking speeds lead to higher loading rates (Hunt et al., 2010), as shown with the higher SR/higher speed combinations in this study. Holt’s data, however, showed a steeper increase from PSR to +30% than from -30% to PSR. The data from Part 2 seem to be increasing linearly throughout the range of SR (rather than a unique slope above and below PSR). The EMG data followed the same trend of increasing with increasing SR and walking speed. In order to walk faster, greater muscle activity is needed, as shown in the higher EMG values for higher SR/faster walking speed. EMG data were greater during both the swing and stance phases. The muscle groups observed in this study are traditionally used for knee flexion in early stance, knee extension in late stance, ankle
plantar flexion at pushoff, and toe clearance during swing. When walking at higher velocities and step rates, all of these joint actions must be quicker.

It was found that GRF and EMG values increase with increasing SR, both for constant walking speed and for constant SL. These GRF results may have implications toward unilateral lower-extremity amputee subjects or other populations where SR is affected. Amputee subjects walk with asymmetrical (between limb) SR, with the intact limb having a higher SR. The intact limb of unilateral lower-extremity amputees develops knee OA at a higher rate than non-amputees (Melzer et al., 2001). The greater GRFv and GRFv LR (both of which are risk factors of knee OA) observed with higher SR in this study may help to explain the higher knee OA rate seen in the intact limb of unilateral amputees. Furthermore, higher muscle forces, as indicated by higher EMG signals, increase bone loading at the knee joint and exacerbate the development of knee OA. It should also be noted that large decreases in SR also result in higher GRF values (for a constant walking speed). This implies that there is an optimal range of SR and SL where forces can be minimized during walking, and any form of intervention should take heed to not stray too far from the PSR.
Chapter 5

CONCLUSION

It was hypothesized that GRF and muscle activation would be minimized at the PSR when walking speed was constant and at the lowest SR when SL was held constant. The results from this study support these hypotheses. The data trends show that the GRF values are minimized at the PSR for a constant walking speed, and follow a U-shaped trend with SR. The findings from this study support published data and extend our understanding of the relationship between GRF and SR. Furthermore, these results offer clinical insight into the potential contributors to increased limb loading seen in amputee subjects. Future work could use similar methods but instead have some sort of manipulation to create asymmetrical gait. By having subjects walk with different step rates for each limb, a better understanding of the underlying mechanisms of amputee gait may be gained.
REFERENCES


Appendix A

LITERATURE REVIEW

*The Effects of Step Rate*

People have a tendency to adopt a certain step frequency and step length walking pattern that minimizes their metabolic cost of walking (Holt et al., 1991). There is physiological evidence to suggest that there are reasons why people walk at their preferred gait. Therefore, it is reasonable to postulate that the body may optimize other criteria besides metabolic cost, such as the forces acting on the body or muscle activation.

Several studies have investigated the preferred step rate (PSR) and why people choose to adopt a specific step rate. Studies focus on the question of what is optimized at the PSR. Holt et al, 1991 found that at the PSR oxygen consumption (analogous to metabolic cost) is minimized, but that mechanical power is not. Umberger and Martin (2007) confirmed that net metabolic rate was minimized at the PSR; furthermore they found that power generation and absorption were minimized at a step rate lower than the PSR by 11-12% but that mechanical efficiency was maximized at a step rate 8% higher than the PSR. This study confirmed that metabolic cost is minimized at the PSR and suggested that there is a compromise of optimizing mechanical power and efficiency near the PSR.
While metabolic cost is minimized at the nominal step rate, the rises in metabolic cost from step rate are not similar for smaller and larger step frequencies. A lower step frequency results in a more dramatic increase in metabolic cost than an equivalent higher step frequency from the preferred case (Holt et al., 1991). A possible explanation for this is that at the lower step frequency (and longer step length), the subjects experience more instability, which requires the recruitment of additional muscle groups to avoid falling (Holt et al., 1991). Umberger and Martin, 2007 agreed with this explanation with qualitative descriptions of the subjects having greater balance demands in the frontal plane at lower stride rates.

There have been several studies that have looked at how ground reaction forces (GRF) change with different step rates (Martin and Marsh, 1992, Russell et al., 2010, Soames and Richardson, 1985). Martin and Marsh (1992) found significant differences in contact time, braking and propulsive peak forces, and impulse at the +10% step rate condition, which was the largest manipulation. There were no significant changes in vertical GRF, most likely due to the minor manipulations of step rate. Russell et al (2010) also did not find significant differences in vertical GRF with increasing step rate, but the data followed a trend toward reducing the GRF. They decreased step length by 15% (and therefore increased step rate) in obese and healthy-weight subjects, in the hopes of increasing metabolic cost while reducing impact. The manipulations may have been too minor in order to see significant changes in the vertical GRF impact and knee adduction moments, as the researchers suggested. Also, the subjects walked at a self-selected speed which may have skewed the data, since
walking slower results in lower moments and impact forces. By using larger manipulations of step rate, significant changes in vertical GRF may be found. Soames and Richardson (1985) found that both step length and step rate have significant effects on the GRF at heel-strike. The GRF in the vertical direction increased with the highest step rate and the peak forces were only significant between the longest and shortest step lengths, 50% and 100% of leg length, respectively. The lowest step rate in this study was 42 strides/min and the highest was 62 strides/min. These results further support the idea that the manipulation in stride rate needs to be pronounced in order to see significant changes.

People walk at a PSR in order to minimize metabolic cost and there are also data to support that there is a compromise of minimizing mechanical power and maximizing efficiency (Holt et al., 1991, Umberger and Martin, 2007). Some studies have suggested that GRF may also be minimized at the PSR, but no significant differences were found. While expected trends were observed, the manipulations may have been too minor in order to see significance.

Knee Osteoarthritis

Arthritis is a widespread problem for the U.S.; 21.6% of adults are afflicted, with osteoarthritis (OA) being the most prevalent type of arthritis (Hootman et al., 2006). There are several factors that researchers suggest could lead up to or exacerbate the development of knee OA. One of which is an increased internal knee abduction moment (Amin et al., 2004, Zhao et al., 2007). The abduction moment is substantial
and accounts for a large portion of the loading at the knee, which carries 60-70% of the weight-bearing forces during walking (Andriacchi, 2004). With greater ground reaction forces, the abduction moment increases, and consequently the medial loading of the knee also increases.

People who have suffered a unilateral transtibial amputation are at an even greater risk of developing knee OA in their intact leg, and have a greater than 65% reported incidence of OA, notably higher than the healthy population (Melzer et al., 2001). An amputee subject’s intact limb experiences greater forces and develops increased bone mineral density (BMD) on the proximal tibia (Burger et al., 1996, Burke et al., 1978, Royer and Koenig, 2005), both of which are potential precursors of knee OA. The imbalance of forces is a result of the asymmetrical gait patterns that amputee subjects exhibit (Dingwell et al., 1996, Sanderson and Martin, 1997). The increased stress resulting from the higher GRF acting on the intact limb of amputee subjects contributes to the degradation of cartilage, tendons, and ligaments, (Hamill et al., 1999). With higher GRF, joint moments, or muscle activity, there is an increase in bone loading at the knee joint, which raises the risk of developing OA at that joint.

Pathological Gait

Unilateral, lower-extremity amputee subjects favor their intact limb resulting in a greater stance time on that side, which leads to kinetic and spatiotemporal asymmetries. It has been shown that there is a decrease in stance time and increase in swing time for the prosthetic limb (Sanderson and Martin, 1997, Isakov et al. 1996) as
subjects are trying to unload their prosthesis because of issues of pain, instability, weakness, or loss of sensory feedback. They step onto the more stable, intact limb more quickly, which results in a higher step rate for the intact limb. The intact limb has an increased stance time and decreased swing time, so the prosthetic side swings through slowly and has a smaller step rate.

Persons with a unilateral amputation experience greater peak GRF in their intact limb as compared to their prosthetic limb (Sanderson and Martin, 1997, Beyaert et al 2008, Nolan et al., 2003). Sanderson and Martin, 1997 found a 1% increase in the intact limb vertical ground reaction force (GRFv) compared to the prosthetic limb. The intact limb’s GRFv were 3% higher than healthy control subjects. While these were minor increases, other studies found significant results when comparing GRF. Beyaert et al. (2008) found the GRF of the intact limb to be 14% higher compared to prosthetic limbs, and 9% higher than the control limbs. Nolan et al. (2003) had similar findings, and also found that the differences in GRF values increased with increasing walking speed.

The internal knee abduction moment is also found to be significantly greater in the intact limb than on the prosthetic side (Royer and Wasilewski, 2006). The knee abduction moment of the intact limb showed a 46% increase from the prosthetic side and a 17% increase from normal subjects. Increases in the knee abduction moment during walking lead to an increase in the compressive forces acting on the medial articular surface of the knee (Hootman et al., 2006, Andriacchi, 2004, Lewek et al., 2005, Zhao et al., 2006). These larger compressive forces put additional stresses on the
articular cartilage and could accelerate the development of physiological problems such as knee OA.

Because of the problems that develop with the healthy limb in pathological gait, it is important to understand the underlying mechanisms of the increased forces that the intact limb experiences. Amputee and stroke patients or anybody that has a condition that affects the lower limbs unilaterally (such as an injury or wearing a brace) tend to favor the good or intact limb and often show higher forces in this limb.

Summary

While it is established that walking at step rates outside the PSR increases metabolic cost, there is inadequate data to show if other factors are affected. If a faster step rate results in increases in forces, this would help to explain why the intact limb of unilateral lower-extremity amputee subjects experiences increased loading. Increases in loading at the knee are especially detrimental to amputee subjects because they will often develop knee osteoarthritis (OA) in their intact limb (Melzer et al., 2001).
REFERENCES


### Appendix B

#### DATA TABLES

**Table B.1:** Statistical results for GRF for constant walking speed (Part 1).

<table>
<thead>
<tr>
<th>SR Condition</th>
<th>GRFv (BW)</th>
<th>GRFap (BW)</th>
<th>GRFv LR (BW/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>-30%</td>
<td>1.29 (0.11)&lt;sup&gt;b&lt;/sup&gt;</td>
<td>0.31 (0.03)&lt;sup&gt;b&lt;/sup&gt;</td>
<td>9.82 (2.23)&lt;sup&gt;b&lt;/sup&gt;</td>
</tr>
<tr>
<td>-20%</td>
<td>1.22 (0.08)&lt;sup&gt;ab&lt;/sup&gt;</td>
<td>0.27 (0.02)&lt;sup&gt;ab&lt;/sup&gt;</td>
<td>9.21 (1.94)&lt;sup&gt;b&lt;/sup&gt;</td>
</tr>
<tr>
<td>-10%</td>
<td>1.18 (0.05)&lt;sup&gt;a&lt;/sup&gt;</td>
<td>0.25 (0.02)&lt;sup&gt;ab&lt;/sup&gt;</td>
<td>8.26 (1.27)&lt;sup&gt;ab&lt;/sup&gt;</td>
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<tr>
<td>Preferred</td>
<td>1.17 (0.05)</td>
<td>0.22 (0.01)&lt;sup&gt;a&lt;/sup&gt;</td>
<td>8.59 (1.19)&lt;sup&gt;a&lt;/sup&gt;</td>
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<td>+10%</td>
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<td>0.21 (0.02)&lt;sup&gt;ab&lt;/sup&gt;</td>
<td>9.64 (1.80)&lt;sup&gt;ab&lt;/sup&gt;</td>
</tr>
<tr>
<td>+20%</td>
<td>1.20 (0.10)</td>
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<td>10.71 (1.79)&lt;sup&gt;ab&lt;/sup&gt;</td>
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<td>+30%</td>
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<td>12.07 (1.78)&lt;sup&gt;ab&lt;/sup&gt;</td>
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a=sig diff (p<0.01) from previous condition  
b=sig diff (p<0.01) from preferred condition

**Table B.2:** Statistical results for EMGst for constant walking speed (Part 1).

<table>
<thead>
<tr>
<th>SR Condition</th>
<th>TA</th>
<th>VL</th>
<th>BF</th>
<th>MG</th>
<th>SO</th>
</tr>
</thead>
<tbody>
<tr>
<td>-30%</td>
<td>136 (36.0)&lt;sup&gt;b&lt;/sup&gt;</td>
<td>177 (79.2)&lt;sup&gt;b&lt;/sup&gt;</td>
<td>220 (122.1)&lt;sup&gt;b&lt;/sup&gt;</td>
<td>126 (44.0)</td>
<td>154(45.1)&lt;sup&gt;b&lt;/sup&gt;</td>
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<tr>
<td>-20%</td>
<td>119 (26.4)&lt;sup&gt;ab&lt;/sup&gt;</td>
<td>127 (35.3)&lt;sup&gt;ab&lt;/sup&gt;</td>
<td>141 (55.0)&lt;sup&gt;ab&lt;/sup&gt;</td>
<td>115 (26.5)&lt;sup&gt;b&lt;/sup&gt;</td>
<td>130(34.2)&lt;sup&gt;ab&lt;/sup&gt;</td>
</tr>
<tr>
<td>-10%</td>
<td>108 (26.7)</td>
<td>113 (32.0)&lt;sup&gt;a&lt;/sup&gt;</td>
<td>112 (42.9)</td>
<td>103 (27.7)&lt;sup&gt;a&lt;/sup&gt;</td>
<td>109(15.8)&lt;sup&gt;ab&lt;/sup&gt;</td>
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<tr>
<td>Preferred</td>
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<td>100 (4.0)</td>
<td>101 (9.5)</td>
<td>100 (5.1)</td>
<td>99(3.4)&lt;sup&gt;a&lt;/sup&gt;</td>
</tr>
<tr>
<td>+10%</td>
<td>95 (19.9)</td>
<td>102 (19.3)</td>
<td>101 (36.2)</td>
<td>105 (19.8)</td>
<td>98(16.5)</td>
</tr>
<tr>
<td>+20%</td>
<td>93 (18.3)&lt;sup&gt;ab&lt;/sup&gt;</td>
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<td>112 (54.4)</td>
<td>106 (32.7)</td>
<td>99(21.6)</td>
</tr>
<tr>
<td>+30%</td>
<td>98 (42.7)</td>
<td>116 (48.4)</td>
<td>125 (61.7)</td>
<td>106 (28.4)</td>
<td>100(18.7)</td>
</tr>
</tbody>
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a=sig diff (p<0.05) from previous condition  
b=sig diff (p<0.05) from preferred condition
Table B.3: Statistical results for EMGsw for constant walking speed (Part 1).

<table>
<thead>
<tr>
<th>SR Condition</th>
<th>TA</th>
<th>VL</th>
<th>BF</th>
<th>MG</th>
<th>SO</th>
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</thead>
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<tr>
<td>-30%</td>
<td>122 (36.4)\textsuperscript{b}</td>
<td>132 (93.4)</td>
<td>103 (50.2)</td>
<td>100 (60.6)</td>
<td>131 (73.5)</td>
</tr>
<tr>
<td>-20%</td>
<td>116 (28.9)\textsuperscript{b}</td>
<td>106 (36.1)</td>
<td>89 (21.9)</td>
<td>81 (39.4)\textsuperscript{a}</td>
<td>113 (45.9)</td>
</tr>
<tr>
<td>-10%</td>
<td>105 (19.8)</td>
<td>95 (34.1)</td>
<td>90 (20.8)</td>
<td>95 (59.8)</td>
<td>107 (34.2)</td>
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<tr>
<td>Preferred</td>
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<td>100 (6.8)</td>
<td>98 (6.5)</td>
<td>97 (8.8)</td>
<td>102 (9.6)</td>
</tr>
<tr>
<td>+10%</td>
<td>99 (14.9)</td>
<td>118 (31.5)\textsuperscript{a}</td>
<td>120 (28.9)\textsuperscript{ab}</td>
<td>139 (105.1)</td>
<td>107 (24.0)</td>
</tr>
<tr>
<td>+20%</td>
<td>102 (14.8)</td>
<td>132 (39.1)\textsuperscript{b}</td>
<td>127 (39.5)\textsuperscript{b}</td>
<td>137 (42.9)\textsuperscript{b}</td>
<td>115 (36.7)</td>
</tr>
<tr>
<td>+30%</td>
<td>104 (30.1)</td>
<td>150 (91.2)\textsuperscript{b}</td>
<td>139 (45.2)\textsuperscript{b}</td>
<td>152 (109.0)\textsuperscript{b}</td>
<td>125 (55.5)\textsuperscript{b}</td>
</tr>
</tbody>
</table>

\textsuperscript{a}=sig diff (p<0.05) from previous condition  
\textsuperscript{b}=sig diff (p<0.05) from preferred condition

Table B.4: Statistical results for GRF for constant step length (Part 2).

<table>
<thead>
<tr>
<th>SR Condition</th>
<th>GRFv (BW)</th>
<th>GRFap (BW)</th>
<th>GRFv LR (BW/s)</th>
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<td>-30%</td>
<td>1.06 (0.04)\textsuperscript{b}</td>
<td>0.19 (0.02)\textsuperscript{b}</td>
<td>4.68 (1.19)\textsuperscript{b}</td>
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<td>-20%</td>
<td>1.09 (0.04)\textsuperscript{ab}</td>
<td>0.20 (0.01)\textsuperscript{ab}</td>
<td>5.68 (0.92)\textsuperscript{ab}</td>
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<td>-10%</td>
<td>1.13 (0.04)\textsuperscript{ab}</td>
<td>0.21 (0.01)\textsuperscript{ab}</td>
<td>6.80 (0.77)\textsuperscript{ab}</td>
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<td>Preferred</td>
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<td>+10%</td>
<td>1.22 (0.05)\textsuperscript{ab}</td>
<td>0.24 (0.02)\textsuperscript{ab}</td>
<td>10.17 (1.14)\textsuperscript{ab}</td>
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<tr>
<td>+20%</td>
<td>1.27 (0.05)\textsuperscript{ab}</td>
<td>0.25 (0.02)\textsuperscript{ab}</td>
<td>12.35 (1.24)\textsuperscript{ab}</td>
</tr>
<tr>
<td>+30%</td>
<td>1.32 (0.07)\textsuperscript{ab}</td>
<td>0.25 (0.03)\textsuperscript{ab}</td>
<td>14.75 (1.62)\textsuperscript{ab}</td>
</tr>
</tbody>
</table>

\textsuperscript{a}=sig diff (p<0.01) from previous condition  
\textsuperscript{b}=sig diff (p<0.01) from preferred condition

Table B.5: Statistical results for EMGst for constant step length (Part 2).

<table>
<thead>
<tr>
<th>SR Condition</th>
<th>TA</th>
<th>VL</th>
<th>BF</th>
<th>MG</th>
<th>SO</th>
</tr>
</thead>
<tbody>
<tr>
<td>-30%</td>
<td>110 (47.9)</td>
<td>69 (33.2)\textsuperscript{b}</td>
<td>91 (64.5)</td>
<td>90 (26.6)</td>
<td>84 (35.5)\textsuperscript{b}</td>
</tr>
<tr>
<td>-20%</td>
<td>105 (41.6)</td>
<td>78 (29.0)\textsuperscript{b}</td>
<td>79 (43.6)</td>
<td>87 (16.4)\textsuperscript{b}</td>
<td>82 (20.5)\textsuperscript{b}</td>
</tr>
<tr>
<td>-10%</td>
<td>107 (48.2)</td>
<td>91 (24.7)\textsuperscript{a}</td>
<td>92 (53.3)</td>
<td>91 (21.0)</td>
<td>93 (26.5)\textsuperscript{a}</td>
</tr>
<tr>
<td>Preferred</td>
<td>106 (33.4)</td>
<td>100 (15.8)</td>
<td>92 (34.6)</td>
<td>96 (17.0)</td>
<td>99 (17.0)</td>
</tr>
<tr>
<td>+10%</td>
<td>112 (28.9)</td>
<td>120 (34.6)\textsuperscript{ab}</td>
<td>111 (63.4)</td>
<td>105 (17.3)\textsuperscript{ab}</td>
<td>105 (17.4)\textsuperscript{ab}</td>
</tr>
<tr>
<td>+20%</td>
<td>112 (21.0)</td>
<td>133 (46.9)\textsuperscript{b}</td>
<td>131 (84.9)\textsuperscript{ab}</td>
<td>108 (20.1)\textsuperscript{b}</td>
<td>108 (15.8)</td>
</tr>
<tr>
<td>+30%</td>
<td>124 (34.5)\textsuperscript{b}</td>
<td>144 (48.1)\textsuperscript{b}</td>
<td>144 (89.3)\textsuperscript{b}</td>
<td>121 (26.9)\textsuperscript{ab}</td>
<td>124 (22.3)\textsuperscript{ab}</td>
</tr>
</tbody>
</table>

\textsuperscript{a}=sig diff (p<0.05) from previous condition  
\textsuperscript{b}=sig diff (p<0.05) from preferred condition
Table B.6: Statistical results for EMGsw for constant step length (Part 2).

<table>
<thead>
<tr>
<th>SR Condition</th>
<th>TA</th>
<th>VL</th>
<th>BF</th>
<th>MG</th>
<th>SO</th>
</tr>
</thead>
<tbody>
<tr>
<td>-30%</td>
<td>111  (66.4)</td>
<td>65  (44.4)</td>
<td>60  (31.1)</td>
<td>68  (45.7)</td>
<td>82  (43.6)</td>
</tr>
<tr>
<td>-20%</td>
<td>113  (71.2)</td>
<td>67  (30.6)</td>
<td>65  (27.2)</td>
<td>100 (97.1)</td>
<td>79  (30.5)</td>
</tr>
<tr>
<td>-10%</td>
<td>117  (67.9)</td>
<td>82  (33.3)</td>
<td>79  (23.0)</td>
<td>87  (60.6)</td>
<td>87  (23.2)</td>
</tr>
<tr>
<td>Preferred</td>
<td>113  (45.0)</td>
<td>91  (22.0)</td>
<td>94  (22.2)</td>
<td>130 (114.2)</td>
<td>102 (33.6)</td>
</tr>
<tr>
<td>+10%</td>
<td>133  (61.1)</td>
<td>126 (55.7)</td>
<td>114 (45.0)</td>
<td>131 (74.1)</td>
<td>113 (23.0)</td>
</tr>
<tr>
<td>+20%</td>
<td>134  (65.5)</td>
<td>154 (70.6)</td>
<td>132 (55.2)</td>
<td>163 (129.0)</td>
<td>147 (101.4)</td>
</tr>
<tr>
<td>+30%</td>
<td>154  (60.3)</td>
<td>176 (66.6)</td>
<td>149 (56.2)</td>
<td>223 (215.9)</td>
<td>175 (108.0)</td>
</tr>
</tbody>
</table>

a=sig diff (p<0.05) from previous condition
b=sig diff (p<0.05) from preferred condition

Table B.7: Statistical results for knee flexion angle excursion from Part 1 (constant walking speed) and Part 2 (constant step length).

<table>
<thead>
<tr>
<th>SR Condition</th>
<th>Part 1</th>
<th>Part 2</th>
</tr>
</thead>
<tbody>
<tr>
<td>-30%</td>
<td>21.6 (4.7)</td>
<td>9.6 (6.1)</td>
</tr>
<tr>
<td>-20%</td>
<td>18.5 (5.2)</td>
<td>12.7 (6.2)</td>
</tr>
<tr>
<td>-10%</td>
<td>18.5 (4.2)</td>
<td>14.7 (6.3)</td>
</tr>
<tr>
<td>Preferred</td>
<td>16.3 (5.8)</td>
<td>16.9 (5.0)</td>
</tr>
<tr>
<td>10%</td>
<td>14.0 (4.3)</td>
<td>16.9 (3.7)</td>
</tr>
<tr>
<td>20%</td>
<td>11.7 (4.2)</td>
<td>16.0 (3.7)</td>
</tr>
<tr>
<td>30%</td>
<td>9.2 (3.5)</td>
<td>15.0 (3.3)</td>
</tr>
</tbody>
</table>

a=sig diff (p<0.05) from previous condition
b=sig diff (p<0.05) from preferred condition
# Appendix C

**IRB FORM**

Physical Activities Readiness Questionnaire (PAR-Q)

University of Delaware  
PI: Jill Higginson, PhD

<table>
<thead>
<tr>
<th></th>
<th>YES</th>
<th>NO</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Has your doctor ever said that you have heart problems or a heart murmur?</td>
<td>□</td>
<td>□</td>
</tr>
<tr>
<td>2. Do you ever suffer pains in your chest?</td>
<td>□</td>
<td>□</td>
</tr>
<tr>
<td>3. Do you ever pass out, have spells of severe dizziness, or experience a persistent, rapid or irregular heartbeat?</td>
<td>□</td>
<td>□</td>
</tr>
<tr>
<td>4. Has your doctor told you that you currently have high blood pressure for which you are not taking medication (systolic pressure greater than or equal to 160 mmHg. or diastolic pressure greater than or equal to 90 mmHg.)?</td>
<td>□</td>
<td>□</td>
</tr>
<tr>
<td>5. Do you smoke cigarettes?</td>
<td>□</td>
<td>□</td>
</tr>
<tr>
<td>6. Do you have diabetes?</td>
<td>□</td>
<td>□</td>
</tr>
<tr>
<td>7. Do you have a family history of heart disease in parents or siblings prior to the age of 55?</td>
<td>□</td>
<td>□</td>
</tr>
<tr>
<td>8. Has your doctor told you that you currently have high cholesterol for which you are not taking medication?</td>
<td>□</td>
<td>□</td>
</tr>
<tr>
<td>9. Is there any physical reason not mentioned here why you should not perform physical exertion?</td>
<td>□</td>
<td>□</td>
</tr>
</tbody>
</table>

Name: ___________________________________________  Age: ____________

Participant signature: ___________________________  Date: ____________
INFORMED CONSENT FORM

Project: Effect of step length manipulation on healthy walking patterns
Pt. Jill Higginson, Ph.D.

Effect of step length manipulation on healthy walking patterns

Summary
You are invited to participate in a research study conducted by Dr. Jill Higginson and Dr. Todd Royer which will investigate the walking performance of healthy young adults on a treadmill. The purpose of this study is to assess the effect of changing (a) step length, (b) stance posture, (c) treadmill belt speeds, and (d) wearing an ankle brace on walking patterns. We will measure the movement of your legs, ground reaction forces and muscle coordination while you are walking at different speeds and conditions.

Study Description
Participants: If you decide to participate, Dr. Higginson or her research associates will ask you to complete a questionnaire about your health, physical activity level, and medical history. A total of 30 healthy adults between 18 and 40 years of age with no history of musculoskeletal or neurological injury will be eligible for this study.

Protocol: You will be asked to wear shorts, t-shirt and comfortable walking shoes. We will assess your comfortable walking speed in the biomechanics lab located at the University of Delaware. While you are seated, we will attach reflective markers to your legs and upper body using Velcro straps. We may also attach surface electrodes to your legs with tape. We will measure your weight and height. You will be asked to walk on a treadmill at 2.9 mph (equal to 1.3 m/s which is close to your comfortable speed) while step frequency (normal, slow, slower, fast, faster) and knee flexion angle (crouched at normal, slower and faster frequencies) are varied (8 trials at fixed speed). To control step frequency, we will play a metronome and ask you to step on each beat. Next, you will be asked to walk on the treadmill with asymmetric belt speeds (normal/slow, normal/fast, slow/fast, resulting in 3 trials with split belts). Finally we will ask you to wear an ankle brace and repeat a normal walking trial plus two asymmetric trials (3 additional trials). We will collect a total of 14 trials on the treadmill lasting less than 2 minutes each. The movement of your legs and upper body will be recorded using special video cameras which detect reflective objects attached to your body segments. The electrical activity of your leg muscles may be recorded during the walking trials using surface electrodes taped to your skin. We may also collect video for comparison with the motion data. You will have at least two minutes to rest between trials.

Total time: Setup will take approximately 1 hour, the testing session will take approximately 1½ hours, and the total time for your participation will be no longer than 3 hours.

UNIVERSITY OF DELAWARE Participant's initials:
APPROVED BY HSRB
Page 1 of 3

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INFORMED CONSENT FORM
Project: Effect of step length manipulation on healthy walking patterns
PI: Jill Higginson, Ph.D.

Conditions for Participation
You should not participate in this project if you are currently pregnant or have a muscle, bone or nervous system disorder. You will be asked to complete a physical activities readiness questionnaire and will be excluded if your answers suggest you should not physically exert yourself. In addition, we will estimate your body mass index which is a measure of body fat and must be less than 30 for you to participate. Your personal information will remain confidential and will not be released (including any publication) without your written consent. Data obtained from this study will be recorded on a computer and archived indefinitely for future research purposes. If you agree, video acquired during this study may be used as part of educational presentations and we will block out your face so that your identity will not be revealed.

Risks and Benefits
Recording the electrical activity of your muscles using surface electrodes poses very little risk. There may be some minor irritation of the skin around the site of the electrode following the experiment. As with any physical activity, risks during walking include dizziness, discomfort in breathing and heart problems. While walking on the treadmill, a handrail will be within reach. You will receive first aid in the event of injury during this project. If you require additional medical treatment, you will be responsible for the cost.

It is hoped that your participation in this project will improve our understanding of muscle coordination during walking.

Contacts
Further information regarding this study may be obtained from the project director, Dr. Jill Higginson, at telephone number (302) 831-6622. Other questions about your rights as a research subject can be directed to the Chair of the University of Delaware Human Subjects Review Board at (302) 831-2137.

UNIVERSITY OF DELAWARE
APPROVED BY HSRB

[Signature]
DATE: 6/15/09
9:41:10

Participant’s initials: 909
Page 2 of 3
INFORMED CONSENT FORM
Project: Effect of step length manipulation on healthy walking patterns
PI: Jill Higginson, Ph.D.

Subject Consent
I agree to participate in the research study described above. I understand that I may withdraw from this study or the principal investigator may terminate the study at any time.

Name: ____________________________ (please print)
Signature: ____________________________ Date: __________

Video Consent
I will allow video to be taken during data collection. YES / NO Initial: _____

I will allow video to be used as part of educational presentations, provided that my identity is not revealed. YES / NO Initial: _____

UNIVERSITY OF DELAWARE
APPROVED BY HSIRB

[Signature] 6/15/09 - 6/14/10 9:00
DATE

Participant’s initials: ___ Page 3 of ___