PREDICTING THE METABOLIC COST OF LEVEL-GROUND WALKING FROM GAIT SPEED AND PROSTHESIS STIFFNESS

by

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A THESIS

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Title: Predicting the Metabolic Cost of Level-Ground Walking from Gait Speed and Prosthesis Stiffness

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The number of individuals requiring the use of a prosthetic limb continues to rise each year. Use of a prosthesis may help improve health and mobility; however, quality of life (QOL) remains lower for many individuals with limb loss compared to their counterparts with intact limbs. One reason for the lower QOL is due to an increased metabolic demand while walking. Prior research has investigated the effects of prosthesis stiffness and gait speed on metabolic demands. However, few studies explore the effects of these variables while using a variable stiffness foot (VSF) prosthesis. Therefore, the purpose of this study was to use the novel VSF prosthesis to develop a regression model which predicts metabolic demand based on gait speed and prosthesis stiffness settings. This model could be used to help customize prosthesis performance for the user.

Five participants were recruited and three were analyzed for this study. The effect of gait speed on metabolic demand was significant ($p < 0.001$) while the effect of stiffness settings was not statistically significant ($p = 0.199$). There were statistical differences in metabolic cost across the gait speeds in 4 of the 5 stiffness settings ($\alpha = 0.013$). The regression model had an adjusted R^2 value of 0.842 with a normalized root mean square error (NRMSE) of 7.02%. While these differences suggest prosthesis stiffness settings do not have a large effect on metabolic expenditure, subject-specific case studies show that prosthesis settings need to be considered when attempting to lower the metabolic demands of a lower-limb prosthesis.

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Introduction

Background

It is estimated that every year in the United States, over 150,000 individuals undergo lower-limb amputation(s) (LLA) [1]. Historically, limb loss has limited the mobility among persons with LLA. However, prosthesis use can help those individuals retain more mobility. The reported rates of prosthesis use vary widely, with usage rates between 49 [2] to 95 percent [3] being reported for lower-limb amputees. Although lower-limb prostheses are intended to replace or mimic the functionality of the removed limb, research has shown that walking with a prosthesis is associated with greater metabolic demands compared to individuals with intact limbs. Lower-limb prosthesis users also self-select a walking speed that is 7 to 42 percent [4]–[6] slower than healthy able-bodied adults. These slower speeds result in prosthesis users exhibiting similar metabolic cost of gait compared to their able-bodied counter parts when gait speed is not controlled for [7]. Yet, the slower speeds lead to a greater total metabolic cost when walking a comparable distance [7].

Individuals with LLA experience mobility limitations such as decreased balance [8], which decrease quality of life. These mobility deficits may be attenuated by improving user specificity in the design of prosthetic feet. The stiffness of foot prostheses has been previously shown to be associated with comfort and metabolic energy expenditure during gait [9]–[13]. The current passive energy storage and return (ESR) foot-ankle prostheses are designed to store and return energy to the user through compliant springs and other elastic materials [14]. However, traditional ESR designs use a single, fixed stiffness; thus, prosthetic users may require different devices based on the velocity at which they want to move. This contrasts the behavior of a healthy foot-ankle complex, which modulates stiffness in response to varied gait conditions [15]–[17]. A recently developed prototype variable stiffness foot (VSF) prosthesis uses a motor to drive a support fulcrum along the elastic keel to allow for control of the forefoot stiffness [14]. This allows for the VSF to adapt stiffness for various gait conditions, possibly allowing individuals to walk efficiently in a wider range of velocities. This development could lead to improved mobility and, thus, quality of life among individuals with lower-limb loss. While the adaptive capacity of the VSF offers great potential to improve the mobility among prosthesis users, the relationship between prosthesis stiffness, gait speed, and metabolic cost of gait remains unclear.

My thesis project aimed to develop a regression model to predict the metabolic cost of walking with a prosthesis to better the experience for the user by optimizing prosthesis adaptability. This model could be used to improve prosthesis functionality such that the prosthesis more readily mimics the behavior of the biological foot-ankle complex during gait. The ability of a lower-limb prosthesis to function more similarly to an able-bodied ankle would enhance the everyday life of an amputee by allowing them to ambulate under a variety of conditions using the same prosthesis. Current passive prostheses are optimized to walk/run at one speed which disadvantages the user. However, a new system that would allow for multiple stiffnesses could allow for more available speeds. The advancements I hoped to find in my project would allow for one prosthesis to handle a range of conditions from a slow to very brisk walk. For example, this prosthesis would give amputees the ability to progress from a slow walk to a brisk walking pace to catch a bus that they were late to arrive at. A current passive lower-limb prosthesis would not allow this to happen without an extraordinary amount of effort. The results of my study could be used to improve the adaptability of semi-active prostheses such that they adapt to the needs of the user. This allows them to engage in many activities they previously could not participate in, thus, improving their quality of life.

Literature Review

Previous studies have thoroughly investigated the impact of LLA on the energy expenditure of gait, finding that lower-limb amputees expend more energy during walking compared to individuals with intact limbs [18], [19]. However, much of the existing literature is limited by the use of single-stiffness prostheses. Thus, studies cannot simultaneously evaluate the effects of speed and prosthesis stiffness on the energy expenditure of walking.

Before investigating the effects of walking speed and prosthesis stiffness on energy expenditure, it is important to understand the factors that affect the speed at which we walk. Walking speed is defined as the product of stride length (SL) and stride frequency (SF). Thus, a change in speed is achieved by changing either SF or SL. SF, denoted in strides per minute, is the duration of a complete stride cycle which includes the left and right step. SL refers to the distance traveled between successive ground contact of the same foot [20], [21]. Studies show that individuals move at an optimal stride frequency (OSF) to prevent excessive metabolic expenditure. As a result, individuals first adapt their SL, before their SF, to increase their walking speed in an effort to maintain their OSF [22], [23]. During this study, walking speed was always held constant for a given trial because participants walked on treadmills. As a result, to maintain walking speed, if SL increased then SF decreased to compensate during a trial [23]. Therefore, during each speed trial, SF likely dictated SL. However, at the low walking speeds used in this thesis, it is likely subjects adapted to speed changes using both SL and SF [20]. This suggests that neither variable will have a greater effect on energy expenditure than the other as participants tried to maintain OSF and used SL to change their walking speed.

Experimentation by Gailey et al. (1994) and Genin et al. (2008) explored the effect of walking speed on energy expenditure. The work by Genin et al. [24] focused on the effect of varying walking speeds on energy expenditure in lower-limb amputees, including trans-femoral and trans-tibial. The experimental results depicted a loss in the maximum (comfortable) walking speed for both levels of amputation as well as an increase in energy expenditure compared to non-amputee control subjects. For those with trans-femoral amputations, energy expenditure was between 30 to 60% greater than the control group. Trans-tibial amputees (TTAs) experienced an increase in their energy expenditure between 0 to 15% compared to the control group. In contrast, the study by Gailey et al. [18] focused on the energy expenditure at self-selected walking speeds. They found that not only was the self-selected walking speed of TTAs 11% slower than nondisabled controls, but that TTAs VO_2 expenditure was 16% greater. These findings are important as they show that no matter the speed (self-selected pace, maximum pace, etc.) the energy expenditure is increased. This suggests that the increased metabolic cost may be attributed to prosthesis stiffness and design.

This is where the variable stiffness foot (VSF) prosthesis I used in my thesis comes into play. Glanzer and Adamczyk (2018), designed the novel prototype (Figure

1). The VSF was designed with an actuated keel support system to modulate sagittal forefoot stiffness, which allows the prosthesis to adapt to variable gait conditions. The keel is designed as an overhung cantilever beam, the length of which is modulated by a movable support fulcrum (B) powered by an onboard motor. The total length of the beam (L) is 229 mm while the overhung length (*a*) ranges from 78 to 163 mm. Modulating the length of the overhung beam, allows the VSF to exhibit a range of forefoot stiffness values. The forefoot stiffness ranges from 10 to 32 N/mm, which correspond to variations in the supported length (*l*). The ability of the VSF to adapt its stiffness during gait sets the prosthesis apart from the current high-performance passive prostheses.

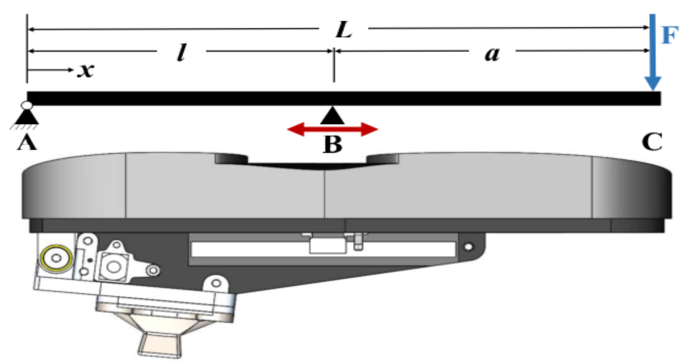

Figure 1: The VSF prosthesis device by Glanzer and Adamczyk (2018)

The VSF modeled as an overhung beam. The diagram shows the keel length (L) attached at A and supported by the fulcrum (B). Overhung length (*a*): L – *l* (supported length).

While a large part of the existing literature incorporates individual aspects of my project, there a few that closely resemble my thesis project. A study by Major et al. [12] explored the effects of stiffness variations on metabolic cost when walking on incline, decline, and level ground surfaces. They found that the lowest dorsiflexion stiffness setting decreased the metabolic cost of participants across level, incline, and decline walking. An additional study found that modulating prosthesis stiffness settings based on gait speed follows a parabolic curve indicating stiffness modulation is important when moving at a speed other than the self-selected pace [9]. This study, however, found that stiffness modulation did not affect the metabolic cost of individuals. These studies provide useful knowledge but allow for my research to explore the interconnection of stiffness, gait speed, and metabolic cost.

Purpose and Hypothesis

The purpose of this study was to determine the relationship between VSF stiffness settings, gait speed, and metabolic cost of gait. These data can be used to develop a mathematical model for predicting optimal stiffness settings to minimize metabolic cost of gait at a range of velocities. Such data could allow prosthesis users to walk efficiently during a variety of conditions and substantially improve their mobility during tasks of daily living. The questions I intended to focus on were:

1) How do prosthesis stiffness settings and gait speed affect the energy required to walk?
2) Can we use prosthesis stiffness settings to increase the spectrum of walking speeds that do not cause a lot of exerted effort?

The ability to walk comfortably at different speeds is an important aspect of daily life. Many lower-limb prosthesis users are limited in their ability to walk comfortably at different speeds due to a prosthesis that is optimized for a single speed and non-adaptable to other speeds. If we can increase the range of comfortable walking speeds for prosthesis users, we improve their overall mobility and quality of life. That is what I hoped to do with my thesis by using the information from my findings to better aid the fitting of prosthesis settings for prosthesis users.

I hypothesized that (1) the regression curve will have a local minimum near the intersection of the 21 N/mm stiffness setting and the 1.0 m/s and 1.2 m/s velocities, and (2) the regression curve will be stiffness and velocity dependent with the velocity-dependent portion being improved by stiffness alterations. My hypothesis of the local minimum is based on the understanding that as you move from an individual's self-selected walking speed to either extreme the more energy it requires to walk. The same can be said as you divert from the middle stiffness setting which balances stiffness and compliance. Understanding how velocity and stiffness impact the metabolic cost of gait for lower-limb prosthesis users will improve the ability to create prostheses that better mimic the biological shank and foot.

Methods

Participants

Five male participants were recruited for this study (Table 1). The inclusion criteria for this study required participants to be injury free for the past three months and be able to walk an extended period of time without assistance. Due to the size of the prototype VSF, recruitment was restricted to participants with men's size 9-10 feet. A signed consent form was obtained from each participant. This study was approved by the Institutional Review Board of the University of Oregon.

Participants	Age (Years)	Height (cm)	Weight (kg)
5	21 ± 1	176.2 ± 5.8	69.9 ± 8.5

Table 1: Participants' anthropometrics. Data are mean ± standard deviation.

Protocol

The research took place at the Bowerman Sports Science Clinic. All study procedures were completed within two hours. In an effort to test our hypothesis prior to addressing the clinical population, this study was tested on healthy able-bodied subjects wearing an amputation simulation device. This device is a restrictive boot designed to mimic the biomechanical and physiological effects of lower-limb loss (e.g. asymmetric gait and increased metabolic cost of gait) and allows for testing of the VSF prosthesis on subjects with intact limbs [14], [25]. The amputation simulation device is a clinically approved walking boot typically worn by individuals with a broken ankle or foot with a prosthesis affixed to the bottom (Figure 2). Subjects wore a modified platform shoe on their opposite foot in order to offset the added height of the amputation simulation device. Initial procedures involved measurements of participants' height and weight.

Then, subjects were given an acclimation period with the boot in which they walk back and forth over ground. Figure 2 illustrates the participant setup during the trials.

Data collections took place on a force plate-instrumented treadmill (Bertec, Columbus, OH). Subjects were asked to walk on the treadmill with the VSF configured under five stiffness settings (15.3 N/mm, 18.6 N/mm, 23.6 N/mm, 28.6 N/mm, 31.9 N/mm) at four speeds (0.8 m/s, 1.0 m/s, 1.2 m/s, 1.4 m/s). The order of the stiffness settings was randomized for each subject with each stiffness having a randomized speed block. Each trial consisted of an initial 210-s adaptation period from rest to the first speed while the following changes in speed had a 120-s adaption period to allow for subjects to biomechanically and physiologically adapt to the new condition [26], [27]. Following metabolic adaption, heart rate and metabolic gas exchange were recorded for 30 seconds. Heart rate was measured using a chest strap (Polar Inc., Bethpage, NY) placed just below the sternum and metabolic energy expenditure was estimated via breath-by-breath metabolic gas exchange (O_2 and CO_2) on a compact, integrated measurement system (ParvoMedics, Salt Lake City, UT). Expired gasses were captured via a mouthpiece supported by wearable head gear. The mouthpiece directs air flow through an N99 filter before entering the measurement chamber via enclosed tubing. After each block of speeds, the treadmill was brought to a stop and the VSF stiffness was manually changed. Ample rest was given between conditions to avoid neuromuscular and cognitive fatigue.

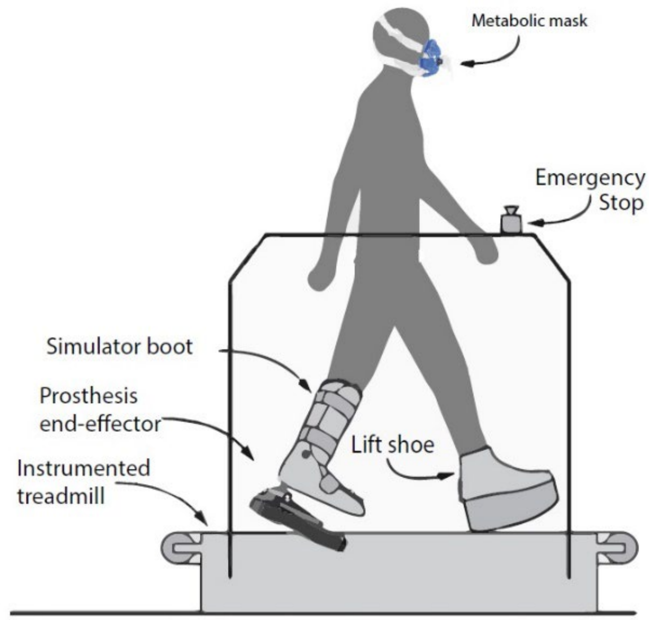

Figure 2: Overview of experimental setup.

This image adapted from Caputo and Collins (2014).

Data Analysis

The final 30-s of data (GRF, heart rate, VO_2, CO_2) were used for all analyses. All raw data were deidentified and processed in Excel and MATLAB (MathWorks, Inc., Natick, MA, USA) using a custom-written script. This script utilized a moving average algorithm to smooth all VO_2 values before exporting them. Processed metabolic gas exchange data, prosthesis stiffness, and walking speed was used to develop a multivariate logistic regression model using the regression function in SPSS (version 27, IBM, Armonk, NY, USA). Due to previous literature stating the value of subject specificity [2], [11], we analyzed both group and a subject-specific case study.

Statistical Analysis

A multifactorial repeated measures ANOVA ($\alpha = 0.05$) was used to analyze the within-trial differences between stiffness settings, walking speed, and stiffness*walking speed on metabolic energy expenditure (VO_2). A Bonferroni correction was applied for pairwise comparisons ($\alpha = 0.013$). All statistical analysis was done using SPSS software.

Results

Group Data

Of the five subjects that participated, only three completed the full experimental protocol because of equipment failures during collections. As such, n = 3 for all analyses. An increase in walking speed was associated with an increase in metabolic expenditure while changes in stiffness settings had little to no effect (Table 2).

Table 2: Effects of stiffness settings and walking velocity on metabolic expenditure (mL O_2/kg/m) ± standard deviation.

	Velocity (m/s)			
Stiffness (N/mm)	0.8	1.0	1.2	1.4
15.3	11.9 ± 1.0*	13.2 ± 0.9	15.4 ± 0.5	18.6 ± 0.9*
18.6	11.9 ± 1.0*†	13.6 ± 0.4#	15.8 ± 0.7*	19.2 ± 0.6†#
23.6	13.1 ± 1.1*†#	14.4 ± 1.2*‡	16.3 ± 1.7†	19.3 ± 0.8#‡
28.6	13.2 ± 1.6*	15.1 ± 1.2†	16.5 ± 1.3#	20.7 ± 1.3*†#
31.9	12.7 ± 1.0*†	14.3 ± 0.4#	16.0 ± 0.6*‡	19.2 ± 0.9†#‡

*†#‡ denote statistical significance (α = 0.013) of pairwise velocity settings within each given stiffness setting.

Changes in gait speed (m/s) caused significant changes in metabolic expenditure ($p < 0.001$). All participants demonstrated an increase in metabolic expenditure as the gait speed increased (Figure 3). Statistical differences (α = 0.013) between gait speed settings occurred at 4 of 5 stiffness settings.

There was no main effect of stiffness settings (N/mm) on metabolic expenditure (p = 0.199). Qualitatively, at 3 of 4 gait speeds there was an in increase in metabolic expenditure between stiffness settings 2 and 3; however, 2 of the 4 gait speeds exhibited a decrease in metabolic expenditure at the 31.9 N/mm stiffness setting (Figure 4).

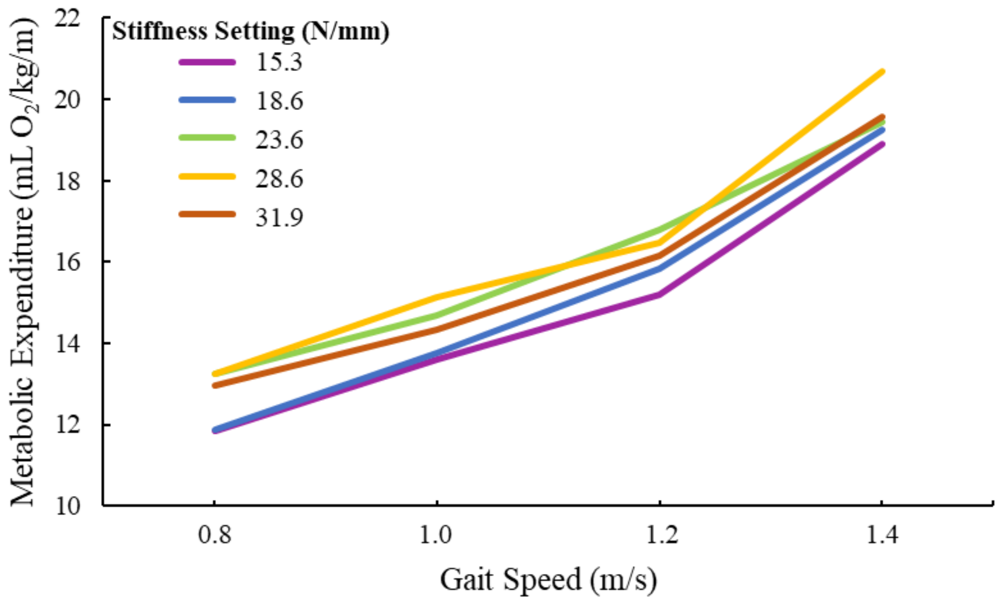

Figure 3: Metabolic expenditure values based on gait speed.

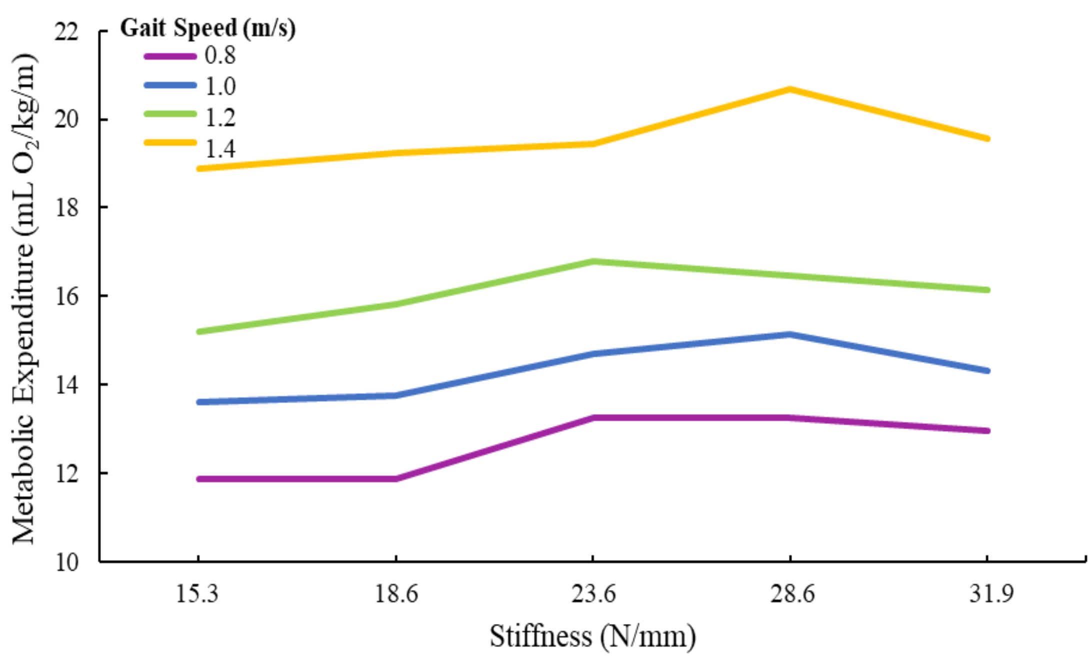

Figure 4: Metabolic expenditure values based on VSF stiffness settings.

Regression Model

The multivariate regression used gait speed and stiffness settings to predict the metabolic expenditure while using the VSF. Based on the shape of the plot, a linear model was used rather than a quadratic function (eq. 1, fig. 5).

$$\text{Equation 1: Metabolic Expenditure } (\text{mL } O_2/\text{kg/m})$$
$$= 2.491 + 0.260 * \text{Stiffness} + 11.115 * \text{GaitSpeed}$$

The regression model had an adjusted R^2 value of 0.842 with a root mean square error (RMSE) of 1.088 mL O_2/kg/m. This represents a 7.02% error based on the normalized root mean square error (NRMSE).

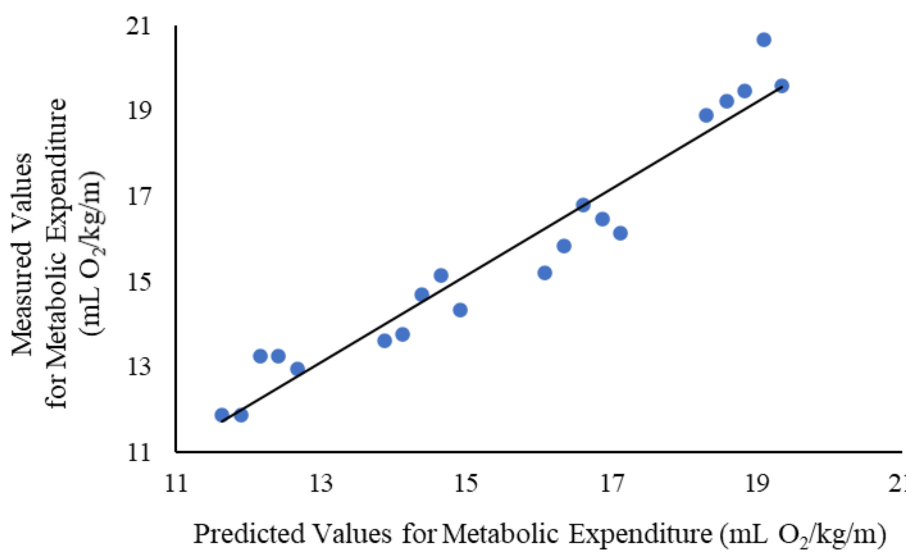

Figure 5: Comparison of the measured and the predicted metabolic expenditure values.

Case Study

Two subjects, referred to as ‘Subject A’ and ‘Subject B’, display the variability of data within prosthetic research. Each subject’s metabolic data differed as a result of both stiffness settings and gait speed settings. The stiffness*gait speed plots reveal the difference in the effects of the dependent variables (Figure 6).

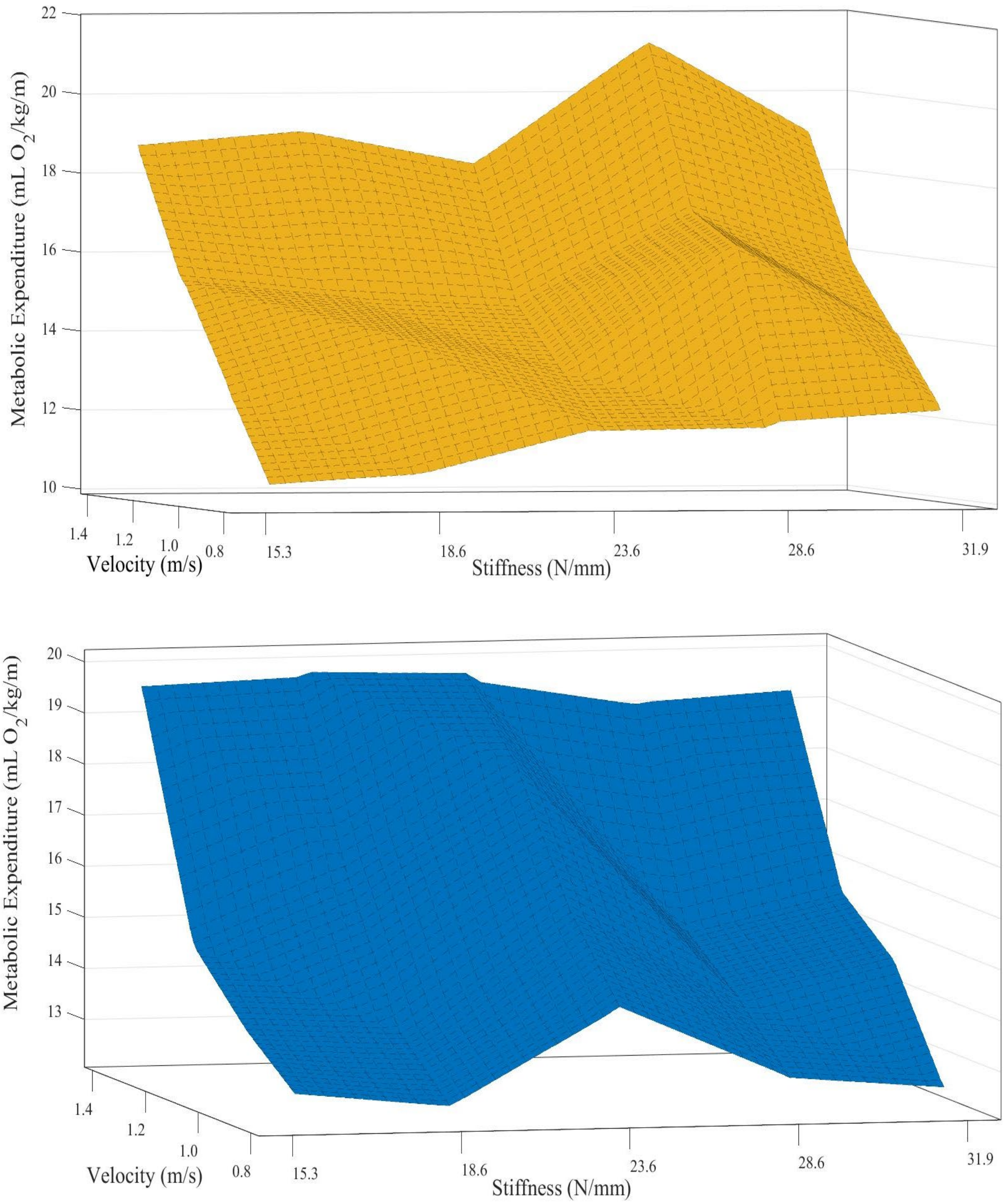

Figure 6: Comparison of the effect of stiffness and gait speed on metabolic expenditure between Subject A (top) and Subject B (bottom).

Discussion

The purpose of this study was to generate a regression model to predict metabolic expenditure using prosthesis stiffness and gait speed to identify the optimal prothesis settings to reduce metabolic expenditure experienced by prosthesis users. We hypothesized that the optimal settings would occur at the intermediate stiffness settings. In addition, we expected metabolic expenditure to increase at the gait speed extremes, creating a three-dimensional, parabolic curve. The results showed a dependence on gait speed with little effect from prosthesis stiffness on metabolic expenditure. Additionally, the regression model followed a linear relationship.

Group Data

Changes in stiffness settings did not have a significant effect on metabolic expenditure (Figure 4). While these results do not match our hypothesis, they represent the ambiguity of this research in the literature. Several recent studies showed similar results regarding the effect of stiffness settings [9], [28]–[31], while an additional study used a prosthetic device that could adapt its stiffness in several locations. This study utilized reduced ankle, toe, and heel stiffnesses while increasing the mid-foot stiffness to optimize metabolic expenditure [10]. Additional studies showed that either low [12], [32], intermediate [33], or high [34] stiffness settings were best to lower energy expenditure. However, many of these studies likely differed from our study as they used several different prosthetic devices during their trials. One study utilized a prosthetic device with a dual-spring mechanism which allowed for modification of plantarflexion and dorsiflexion [12]. Another study worked with several running blade prosthetics, each with a different stiffness setting [32]. A notable component of this study was the use of the VSF device which allowed for a two-fold change in stiffness without swapping out the prosthesis device itself, thereby mitigating confounding variables associated with a foot switching experimental design.

The second observation was the effect of gait speed on metabolic expenditure (Figure 3). This result exhibited the pattern found throughout the literature. Two studies depicted the effect of gait speed on metabolic cost as a parabolic curve [24], [35]. However, when looking at the velocity range used in this study, that portion of the parabolic curve is increasing as velocity increases [24]. Furthermore, the literature depicted a similar relationship for both trans-tibial and trans-femoral amputees [23], [36]. The results of our study do show the influence of the prosthetic simulator boot on metabolic expenditure. Previous studies showed the metabolic expenditure of walking for healthy, able-bodied individuals ranged from 5 to 10 mL O_2/kg/m [37], [38] at similar walking speeds as those used in our study. In contrast, our results ranged from 12 to 20 mL O_2/kg/m (Figure 3). This matches previous literature which stated the use of a prosthesis device increased the user's metabolic cost [18], [24].

Regression Model

The multivariate regression model did not match our hypothesis. Instead of a three-dimensional parabolic shape, the regression model exhibited a linear relationship. This may be attributed to the lack of effect by the prosthesis stiffness and the ranges of the independent variables. Due to the lack of significance from prosthesis stiffness, gait speed ($\beta = 0.909$) impacts the regression more than prosthesis stiffness ($\beta = 0.138$). Therefore, it is understandable that the regression followed the linear shape of the effect of gait speed on metabolic expenditure. Additionally, as stated earlier, previous studies on gait speed followed a parabolic curve when using a larger range of velocities [24], [35]. It is possible that a larger range of velocities in this study would cause the regression trend to appear more parabolic. Further research could expand on the velocities used to validate the regression model. In addition, the exploration of a variable stiffness controller used during the trials may lead to better optimization of metabolic expenditure.

Case Study

The lack of significant effect on metabolic expenditure by prosthesis stiffness called for a closer look at specific subjects. This subject specificity is expected as previous literature has shown that prosthesis stiffness settings, gait characteristics, and subject perception are important factors for optimizing an individual's prosthesis fit [39], [40]. In our study, the increase in metabolic expenditure due to the effect of gait speed was similar between Subject A and Subject B. However, the changes in metabolic expenditure due to prosthesis stiffness was quantitatively different. As seen in Figure 6, at the 23.6 N/mm stiffness, Subject A has a trough while Subject B has a large crest. Vice versa, Subject A has a crest at 28.6 N/mm while Subject B has a trough. Additionally, the subjects differed in their metabolic response during specific walking speeds. At 0.8 m/s (lower portion of the Figure 6 plots), as prosthesis stiffness increases the metabolic expenditure of Subject A gradually increases. Contrarily, besides a large jump in metabolic expenditure at 23.6 N/mm, there was little change in metabolic expenditure for Subject B as stiffness increases. Similar differences are seen at 1.4 m/s (top portion of the Figure 6 plots), where Subject B sees little change in expenditure while Subject A sees large changes. We expect this subject specificity because previous literature shows an important part of the rehabilitative outcome is the fit of a prosthesis and its components [2], [11]. Additionally, the literature has shown that prosthesis model and fit plays a large role in optimizing the metabolic expenditure of the user [28]. While it is known that prosthetic socket fit leads to better performance, a study by Pezzin et al. [3] found the lowest level of satisfaction was related to prosthesis fit, specifically socket fit and alignment. This only further supports the priority of prosthesis fit to the overall functional outcome of a prosthesis, a subject specific issue.

There are a few reasons that these subjects differed in their optimal stiffness setting. The first is based on the subjects' activity level and lower-limb muscle mass. The prosthetic simulator boot adds an additional, unfamiliar weight to the subjects' lower limb. Depending on each subject's muscle mass, this could cause increased use of hip muscles which would lead to changes in gait and a difference in the optimal stiffness. Additional gait characteristics that could explain the difference in optimal stiffness settings between subjects include SL symmetry and limb loading. Subject A loaded the lateral aspect of the prosthesis during walking. This caused increased torque on the foot and ankle which could be exacerbated by the higher stiffnesses due to low compliance. This could explain why Subject A's optimal stiffness setting was at 15.3 N/mm.

Implications

Our results demonstrate the importance of subject specificity when fitting a user with a prosthesis. Although the group data did not have a main effect, the difference in trends between Subject A and B portray how different individuals respond to various prosthesis stiffness settings. The regression model developed in this study could be applied to the fitting process to better optimize prosthesis stiffness at a desired gait speed. Additionally, the model could be applied to determine the effect of minute stiffness changes at a constant gait speed. If research continues to validate the importance of subject specific prosthesis stiffness settings, clinicians and manufacturers need to factor in the effect of minute stiffness changes. This would allow for better optimization of metabolic expenditure for prosthesis users.

Limitations

There are limitations within my experimental protocol that should be acknowledged. The modified walking boot and the lift shoe are a men's size 10 shoe. This caused the participants within my study to be exclusively men, primarily of similar height and build. Additionally, only 3 subjects were used in the analysis. This limited our statistical power and may have decreased any effects we could have seen. Finally, all participants were not from the amputee population. While this means my sample size does not reflect the population I am studying, I believe the project can still be applied to benefit the lives of prosthesis users due to the normalization of the metabolic data. All of the metabolic data is normalized using each participant's weight. This allows the data to be compared across a variety of individuals. I hope that future research will be able to incorporate women and a larger range of shoe sizes to better represent the overall population.

COVID-19 Pandemic

I would be remiss if I did not mention the impact of the Coronavirus pandemic that affected our lives and my research. As of writing this, I still have not collected all 15 subjects like originally planned even though I started the thesis process back in

February of 2020. The university and state guidelines shut down our research group at the end of March 2020. This shut down continued until September/October 2020. During that time, I could not collect any data which drastically set me back. I received a grant for the summer of 2020 to conduct research but, instead, I spent the summer coding as much as I possibly could to limit time spent on future data analysis. Once research opened again, COVID-19 affected my research in other ways. These included changes to the protocol to allow for new time limits, decreased lab accessibility, and stricter safety protocols in the lab. While COVID-19 did not entirely defeat my thesis, it dramatically stalled the process.

Conclusion

This study attempted to fill the current gap in the knowledge of the effect of prosthesis stiffness and gait speed on metabolic expenditure to better adapt to prosthesis users. Our study reported an increase in metabolic expenditure with increasing gait velocities, with no significant effect by prosthesis stiffness. Our regression model followed a linear relationship with a large emphasis on gait speed. The lack of statistical significance from prosthesis stiffness settings furthers the importance of subject specific fitting of prosthetic devices.

Although many studies have found inconclusive stiffness results, few have used a single prosthesis device to modify stiffness settings. Future studies should look to broaden the range of both stiffness settings and gait velocities to better predict subject energy expenditure. Additionally, future studies should adjust prosthesis stiffness during ambulation to closer emulate the human foot-ankle complex.

Appendix A: Repair Journal

While this thesis journey was much longer than I initially planned for, the added length did allow me to learn much more about the VSF prothesis. Over this last year and a half, the foot broke several times. The first of many repairs occurred after the foot imploded during a pilot test. The nylon tensioner bolt broke which sent pieces flying everywhere (Figure 7). This repair took quite a long time due to shipping delays; however, this did not become an issue as research was shut down at the university due to the pandemic.

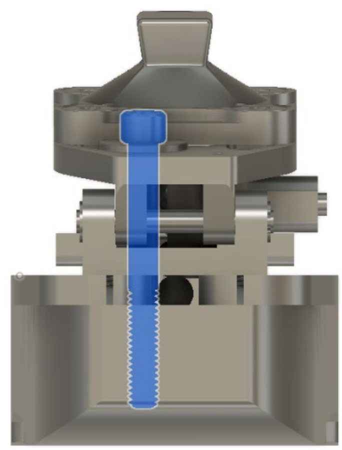

Figure 7: Posterior view of nylon tensioner bolt used to prevent the keel from flopping downward.

The next repair that we needed to fix was a burnt-out motor (Figure 8) and fried Wixel radio transceiver board on the VSF. These two repairs meant the foot also required a full rewiring to make sure all the new parts would work together properly. In addition, we devised a 3D-printed housing (Figure 9) for the on-board Wixel to allow for better protection of the wires.

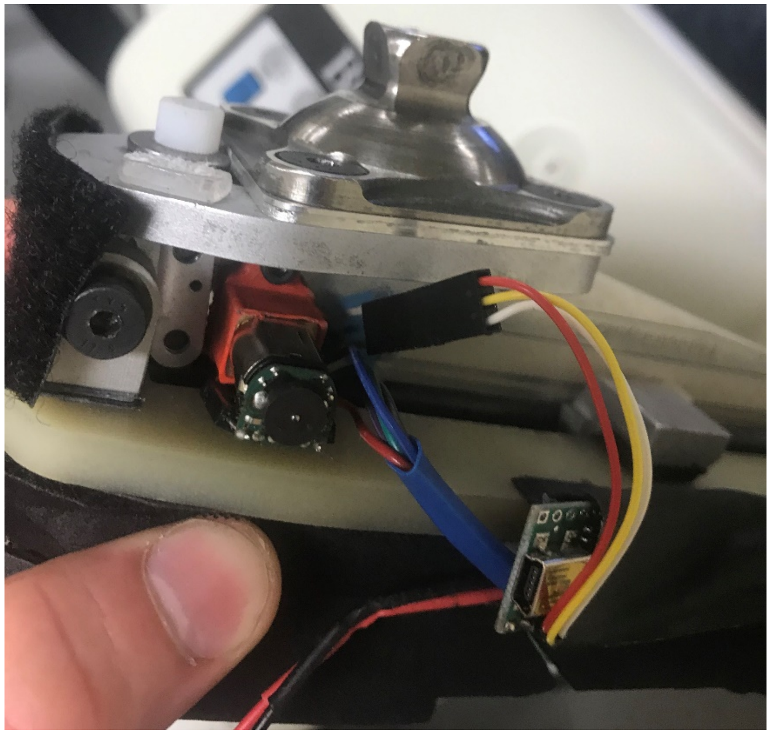

Figure 8: VSF motor which drives the fulcrum to the desired stiffness setting.

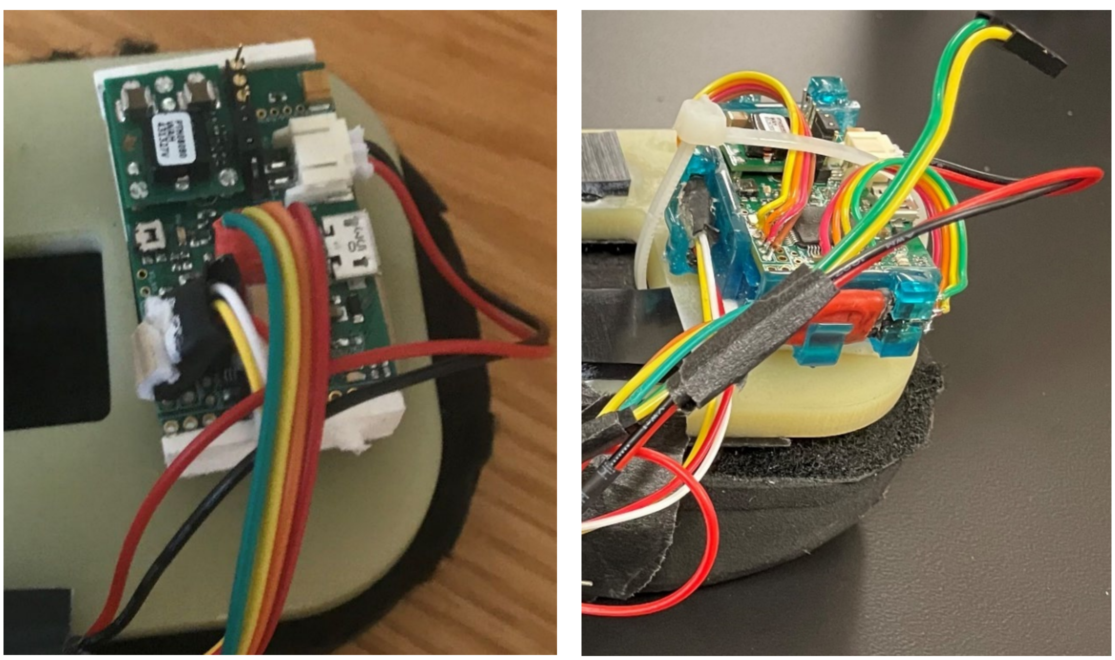

Figure 9: The on-board Wixel in the old housing (left) versus the new housing (right).

The last repair we made to the VSF prosthesis occurred after a subject loaded the lateral aspect of the foot which caused the alignment beam to shift to the right and the keel to snap (Figure 10). This repair required us to replace the entirety of the keel and reconfigure the fitting process to prevent further loading of the lateral aspect (Figure 11). Additionally, we redesigned the fulcrum padding to prevent the padding from falling off. The new design glued the padding along the entire length of the keel rather than on the fulcrum itself.

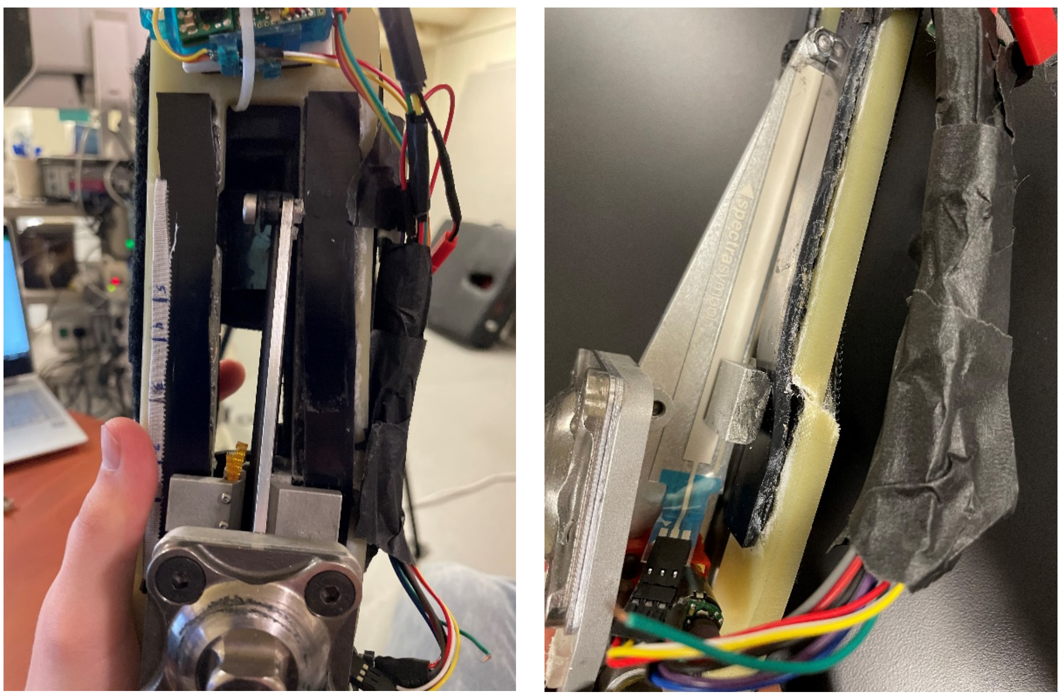

Figure 10: Loading on the lateral aspect of the VSF and the resulting torque caused the alignment beam (left) to shift to the right and the keel (right) to crack.

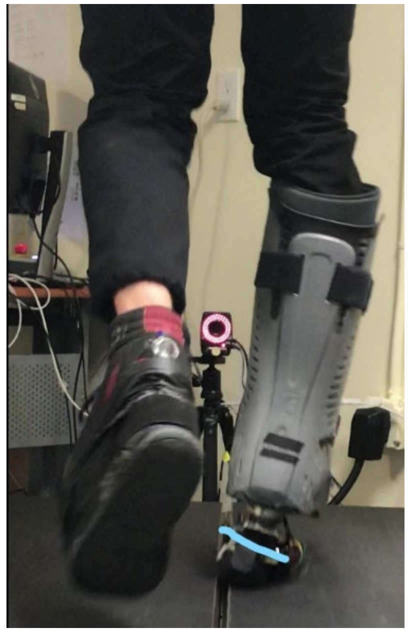

Figure 11: Example of a subject loading the lateral aspect of the VSF prosthesis.

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