Enhancing Human Metabolic Economy in Stair Climbing via an Elastic Crutch Mechanism

by

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Abstract

Crutching provides a significant increase in mobility for those with limited walking ability. While level ground walking with crutches has been studied in many different forms, stair climbing is a more energetically taxing activity and the upper arm and shoulder strength required is not always available in weaker or severely injured patients. We posit that the introduction of parallel springs spanning the elbow joint will improve the crutching experience by helping patients attain a metabolic reduction compared to unassisted locomotion.

Here, we present a foundation for achieving metabolic reduction with joint-spanning elastic elements. Our approach includes three parts. First, we present an augmented crutch design with an elbow spring that can be modified with different stiffnesses. Second, we put forth a clinical testing protocol in which we measure metabolic economy via the pulmonary gas exchange technique ($\dot{V}_{O2avg}$). Simultaneously recording electromyographic (EMG) signals from the primary active muscles provides a neuromuscular interpretation of the crutching activity not captured by the black-box metabolic techniques. We complete the picture by modeling the energetics of the effective elbow muscle by incorporating empirical measurements of maximum angular velocity achieved under a range of isotonic conditions.

The metabolic data exhibits trends consistent with our hypothesis of metabolic reduction; although, more subjects are needed to confirm these results. All subjects reported a feeling of augmentation at the optimal stiffness condition. An analysis of the EMG results show a clear transition in muscle usage patterns from a triceps only power stroke to a combined usage of both triceps and biceps. Where the triceps are maximally active during the non-augmented state, as stiffness increases the biceps become more active and the total activation level drops, suggesting the this shift is at least partially responsible for the observed metabolic reduction. While the model correctly predicts the relative shape of the observed curve, the optimal stiffness predictions are higher than their empirical equivalents. This is most likely due to the extra help the triceps muscles received from active stabilization and power muscles not considered in the model. With a more complete muscular picture one could begin to construct an accurate method of prediction and tuning of optimal stiffness.

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Chapter 1

Introduction

1.1 Literature Review

Crutches have long been prescribed as mobility aids for musculoskeletal and neurological pathologies resulting in limiting walking ability. Beneficiaries of these ambulatory aids include patients with fractures, amputations, joint replacements, paraplegias, and the elderly. Functionally, crutches must support the body during locomotion by transmitting whole or partial body weight (up to 80% [22]) through the hand to assist in the stability, support, and propulsion of gait and to increase maneuverability.

1.1.1 Crutch designs

The most common crutch styles include underarm crutches (also known as axillary crutches), Lofstrand crutches (also known as forearm crutches), triceps crutches, and platform crutches, Figure 1-1. Forearm crutches are often used for those with chronic or permanent injuries or ailments as they provide in general more mobility at the risk of being less stable [13]. Underarm crutches are most popular in the United States for short-term injuries. However, Lofstrand crutches have been more highly prescribed in recent years for both long-term and short-term injuries due to the greater maneuverability they provide for both the user's arms and legs. Most of these designs also include generous padding on the handles to cushion compressive stress on the palm and a rubber tip at the base to increase friction and stability. In addition to the most popular styles, interesting iterations on these themes exist including rocker-crutches, crutches with a base spring to cushion the crutch stance phase, new handle designs, harness crutches, new materials including compliant composite and molded glass-reinforced fiber, Figure 1-1. To date, none of these experimental devices have proven commercially viable.

1.1.2 Crutching Gait

The crutching gait in general has been studied both kinetically [34][23][35] and kinematically [25][27] in a variety of settings. Crutches have been evaluated metabolically with pulmonary gas exchange ($\dot{V}_{O_2avg}$) and blood pressure metrics to ensure proper fit, compare crutch styles, characterize speed, and analyze crutching gaits with respect to normal ambulation [28][31][15][11]. For biomechanics studies in particular, illustrating metabolic rate via $\dot{V}_{O_2 avg}$ intake is a well-understood method by which the volume of O_2 inhaled used to calculate to energy intake. All of these studies show that crutches are generally much less efficient than normal ambulation while remaining relatively inconclusive on the most efficient crutching style.

While some studies have compared the energetics of stair climbing between elbow and axillary crutches [5], little work has been done on improving the crutching experience up stairs. With underarm crutches stair-climbing is nearly impossible, most physicians recommend to hold the crutches to the side and hop up the stairs [9]. With Lofstrand crutches, stair climbing is possible as there is no handle in the way; however, the act of stair climbing takes a large amount of energy and the upper arm and shoulder strength required to complete this task is not always a suitable option for weaker or severely injured patients. As crutching provides a significant increase in mobility and independence compared to self-wheeled or powered wheelchairs and walkers and is less expensive than other similar prostheses and orthotics (especially in tight spaces and over non-level and uneven terrain), it is important to pursue opportunities that enhance the ease of the crutching experience.

1.1.3 Elastic Elements in Crutch Design

Series elastic elements have been previously suggested as dampers and control elements in crutch design [19][26]. Series devices, while similar, do not directly aid muscles, but rather require the human body to adapt to new movement patterns such that muscles can operate in a more optimal regime [33]. Parallel elastic elements have been described previously for single joints and multiple joints. One paper suggests that by optimizing parallel elastic elements over a joint, that one can enhance human endurance as measured by time to muscular exhaustion [12]. A similar patent by the same author suggests designing crutches with elbow springs for the purpose of optimization [13]. In the case of [13], cycles until exhaustion establish a good metric as the measured activity itself (pull-ups) was highly fatiguing. In the case of crutching, exhaustive fatigue is a much more elusive concept as stabilizing muscles fail first, and therefore a metabolic metric presents a more practical approach. This thesis works on the optimization of this particular principle.

1.2 Thesis Objectives

The objective of this thesis was to design an experiment to test the following hypotheses:

1. Joints can be tuned to reduce metabolic cost of operation by the addition of parallel spring elements.

2. By treating the set of muscles working over a joint (in this case the elbow joint), as a single "effective muscle", one can make predictions about the parameters

which can induce this optimal energetic minimum.

3. These principles can be used to design an elastic crutch mechanism which reduces the metabolic cost of stair-climbing.

1.3 Chapter Summary

Chapter 1 presents a literature review highlighting previous research and relevant design concepts.

Chapter 2 compiles a list of basic information that would be useful in understanding the scope of the problem including: the basics of the stair-climbing crutching gait and usage patterns, information on a commonly used muscle model and characteristics thereof, a discussion of the proper metrics by which to measure biomechanical advantage, and finally a short list of functional requirements when designing rehabilitative and experimental devices.

Staring with **Chapter 3**, we delve into the experimental methods including device design, data collection tools, subject recruitment, and experimental procedure.

Chapter 4 Continues on this theme by discussing the methods of data analysis for each set of experiments as well as the details of the overarching arm model used to inform experiment design and data processing methods.

Chapter 5 presents all experimental results including statistical analyses and data fitting procedures.

Finally **Chapter 6** presents the discussion of results and final conclusions.

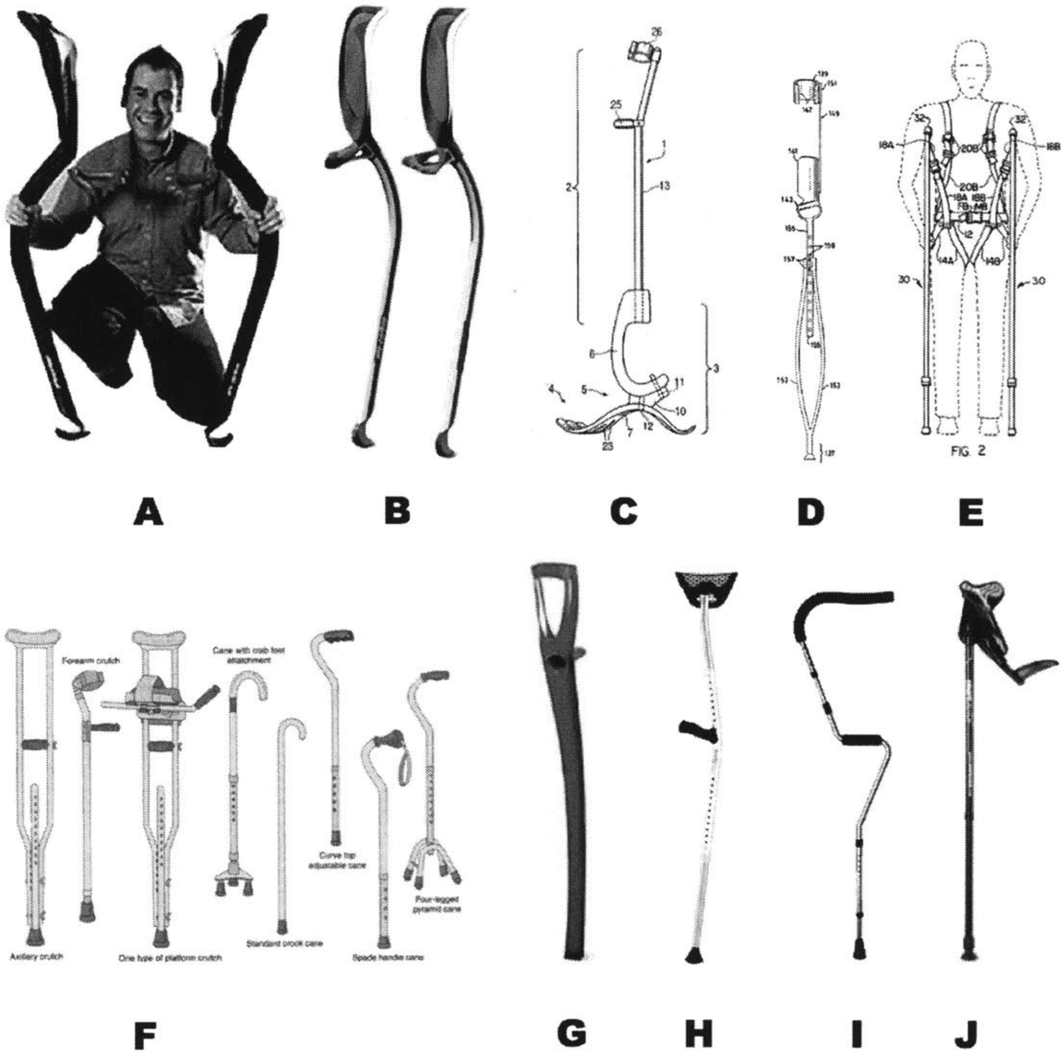


Figure 1-1: Common and experimental crutch desings. [A-B] "Pro-crutch" alternative forearm crutch design, Royal College of Art. [10]. [C] Forearm crutch with spring-foot [32]. [D] Forearm crutch with lower and upper segment compliance [13]. [E] Harness crutches [24]. [F] Crutch designs in common use [22]. [G] Stackable crutches made from molded glass-reinforced plastic, qed Design [3] . [H-I] Crutch with flexible underarm pad [2]. [J] "SmartCrutch" ergonomic arm rest design[4]

Chapter 2

Background

Biomechanics provides a framework for studying biological systems with the tools of mechanical engineering. In general, scientists attempt to come up with simple models for how the body works, and validate them with quantitative data from energetics, structural mechanics, and dynamics.

2.1 Crutching

Human walking is periodic, and therefore may be analyzed using a representative period, called the gait cycle. The gait cycle can be further broken into two phases: stance and swing. The first phase of the gait cycle is the stance phase which can be defined as the period of time when the foot (or crutch) is on the ground. In contrast, swing phase is defined as the period of time when the foot (or crutch) is off the ground. As crutching by definition requires the use of human legs and another assistive device, crutching gaits therefore require one to consider overlapping stance and swing periods for both crutches and the lower limbs. Particularly interesting to note, is that during crutch stance, a person on crutches behaves like inverted pendulum, in other words, the system is dynamically unstable. During this period, the user will often lift both feet off the floor leaving themselves vulnerable to instability and falls. The elevation component in stair climbing only exacerbates these issues. While level ground crutch walking and unassisted stair climbing have been studied in many different forms [22], less guidelines are available when looking at stair climbing in crutches.

2.1.1 Stair-climbing Gait

When discussing gait patterns requiring body elevation, a few general guidelines apply[30]:

1. When climbing upwards, patients are instructed to lead with the unaffected leg, followed by the affected leg and gait aid together.

2. When climbing downwards, patients are instructed to lead with the affected leg and gait aid and follow with the unaffected leg.

In practice, stair climbing presents a challenge for all crutch users and significant gait variation exists among users. For those with underarm crutches, stair climbing by simultaneously loading both crutches is nearly impossible as the frame's axillary portion obstructs elbow flexion. Instead, clinicians recommend hopping up the stairs using a variety of creative load bearing techniques or sliding up the stairs while sitting using ones hands to push up the body one stair at a time. [9]

For patients with forearm crutches, the gait is less restricted. Starting from a double stance phase [1], the patient is instructed to flex the elbow, placing the crutch on the next stair (crutch swing, leg stance). Next, the patient extends his or her arms, lifting the body off the previous stair (crutch stance, leg swing) and up to the next, ending in a second double stance position. However, this gait requires significant arm strength to maintain.

It is important to point out that similar to the increased leg requirements in unaided stair climbing compared with level ground walking, stair climbing with crutches requires the application of larger forces and a greater range of motion from the arms than the level ground equivalent.

2.1.2 Crutching Energetics

The literature presents conflicting evidence on whether the type of crutch (ex. underarm vs. forearm) makes a significant difference in energy usage during elevation activities. In stair-crutching, the major sources of energy expenditure are lifting the body against gravity and stabilizing the body over the crutch posts.

In walking and other forms of locomotion, a commonly used metric is the dimensionless specific cost of transport: the energy expended per unit distance walked, normalized by subject weight.

2.2 Muscles

Whether we move by using our hands or feet, the capabilities of available muscles motivate how we shift weight, change direction, and locomote. In the case of crutches, the muscles of the upper arm and shoulder are of prime importance.

2.2.1 Musculoskeletal Model for Upper Extremities

The Biceps bracii is a two headed muscle located on the front of the upper arm, Figure 2-1 . Its chief job is to flex and supinate the humeronuclear joint (elbow). Its antagonist is the Triceps bracii muscle. A three-headed muscle that lies along the dorsal side of the upper humerus, the triceps primarily functions as the great extensor of the forearm. This capable agonist-antagonist pair enables the joint to operate simultaneously extremely fast and at high torque. Maximum accelerations of the elbow joint have been recorded between 35 and $42 rad/s$. Maximal torque production in both concentric and eccentric directions peaks around $35 - 40 Nm$

[1] In this case, double stance refers to both crutches and legs on the floor

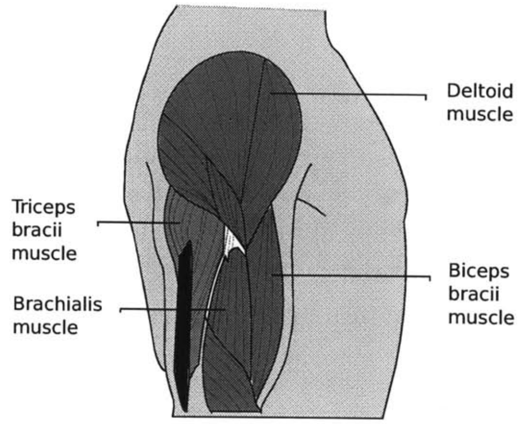


Figure 2-1: Primary arm and shoulder muscles

[17][18] Other muscles of interest include the other main elbow flexor, the Brachialis, and the shoulder muscles especially the Deltoid.

Two muscle tasks are required to execute a crutching elevation gait.

1. *Powered extension and flexion of the elbow joint in the sagittal plane*–Primarily this refers to the biceps and triceps, which are the major power muscles for the elbow joint in this configuration NOTE: Shoulder muscles also work to extend the arm however for the purpose of this experiment we are chiefly focusing on optimizing the elbow joint. The isolated elbow joint is easier to span mechanically and provides a localized benefit that can then be extrapolated for use on other muscles.

2. *Stabilization of the body over the crutch post*– A simple model of a crutch in stance is that of an inverted pendulum. This means that when balancing only on the crutch, the patient dynamically unstable. Stabilization is primarily provided by the the compensatory stiffening of muscles around the shoulder and wrist joints. While this thesis does not attempt to improve stabilization by any quantifiable measure, stabilization is of interest to the extent that it affects the user's ability to move comfortably and safely.

2.2.2 Force-Velocity Characteristic

It has long been known empirically that muscles can shorten more quickly against light loads than against heavy ones. [14] While this phenomenon is partly explained by the inertia of the load and the muscle, it remains a fact that muscles contracting isometrically produce more force than those which are actively shortening. This relationship can be generally described by Hill's equation, a rectangular hyperbola of the form (Figure 2-2):

$$(F + A)(v + b) = (F_0 + a)b \tag{2.1}$$

where F is muscle force, v is the maximum muscle contraction velocity under isotonic load F, the asymptotes are $F = -a$ and $v = -b$, F_0 is the isometric tension (force against which muscle neither shortens nor lengthens, and $v_{max} = \frac{bF_0}{a}$ is the shortening velocity against no load.[2]

This model applies to all human muscles. A normalized form of the equation can be written as:

$$v' = \frac{(1 - F')}{(1 + \frac{F'}{k})} \tag{2.2}$$

where, $v' = \frac{v}{v_{max}}$, $F' = \frac{F}{F0}$, $k = \frac{a}{F_0} = \frac{b}{v_{max}}$

Furthermore, a particular muscle's mechanical power output can be calculated by multiplying the independent and dependent variable in the previous equation to get:

$$Power = F * v \tag{2.3}$$

where the peak is around $0.1 * F_0 v_{max}$ watts

By coming as close as possible to this maximum power point in as many muscles as possible in a system, it is conceivable that one can optimize the system for maximum metabolic efficiency.

2.2.3 Muscle Activation

Muscle are activated by neural signals which through a complex chemical interaction produce contractions in the muscle tissue. One can measure the gross output of these neural signals by a methodology called electromyography. Using this technique, one places electrodes near the muscle site[3] to read the difference in electric potential as muscle contractions occur. In comparing muscle activation at different muscles one can easily perceive any instances of co-activation (where antagonist muscles contract simultaneously). Furthermore, it is important to note that while one can make general statements about the relative activation level for different muscles, electromyography does not provide a complete picture of how the muscle reacts. The neuromuscular system is very rich and complex and there are inherent limitations in reducing all that content into one information signal. Finally, in no way should electromyography be interpreted as a measure of force or torque. While these two signals often follow the same trend, they are by no means consistently correlated.

[2] In this case, v refers to the velocity measured immediately after the muscle is released in a quick release experiment as described in Hill. [14]

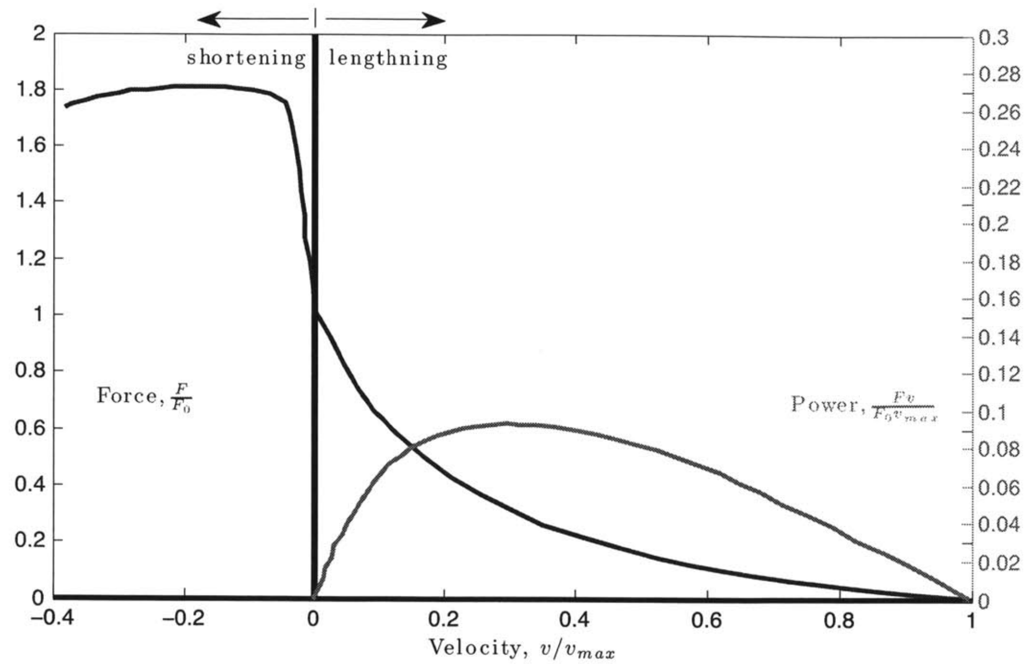


Figure 2-2: A sample normalized force-velocity curve for skeletal muscle.[21] The shortening section of the force-velocity curve can be fit to a rectangular hyperbola. Here, T_0 and T refer to F and F_0 in this document's notation. The power curve for this characteristic is shown as well.

2.2.4 Fatigue vs. Metabolics: Understanding Muscle Usage in the Body

Choosing the proper metric by which to measure biomechanical advantage is nontrivial. Similar studies such as [12] found an optimal biomechanical operating point during heavy exercise by reporting a subject's maximal endurance level as measured by work cycles until exhaustive fatigue. This method works well for highly fatiguing activities, or activities defined as activities that can be performed until exhaustion in the same geometrical configuration using the same muscle set. However, most crutching gaits cannot use this metric as they do not qualify as highly fatiguing. Firstly, the biceps and triceps are rather robust and normal crutching forces are not taxing enough to fatigue them to complete exhaustion in a reasonable amount of time. Additionally, crutching is by definition an unstable activity: the basic dynamic motion mirrors that of an inverted pendulum. Long before muscles fatigue, weaker stabilization muscles activate to produce compensatory motions in the upper arm and shoulder structure to maintain balance. Pilot data has shown that attempts to speed up the fatigue process only increase the compensatory effect. This change in the active muscle complement, confounds and limits the conclusions that can be drawn based on endurance. Finally, endurance as measured by exhaustive fatigue only reflects on a final energetic state while providing no information on the rate at which one approaches fatigue.

Given the stated concerns, metabolic economy data provides a better metric by which to measure optimality. As metabolic data is reported over the course of an experiment, one can better understand the change in fatigue and movement efficiency over time. Erasing the need for complete fatigue reduces the experimental time frame and makes subjects feel safer and more stable. Finally, metabolic data allows us to measure movement efficiency more directly than by relying on pure calculation from mathematical biomechanics models. All these reasons contribute to the decision to use metabolic data as a metric for this experiment.

2.3 Designing Rehabilitative & Experimental Devices

A number of necessary functional requirements emerge when using rehabilitative devices. In approximate order of importance:

1. *Effectiveness* –The device needs to perform its defined purpose, in this case, reducing the metabolic economy of stair climbing.
2. *Comfort* –Devices must not impede anticipated body movement or function. The designer must keep in mind that any pain or discomfort may cause the user to execute non-natural, and often less efficient, motions. For an energetics study this is unacceptable.
3. *Safety* –The device must be safe to use at all times. A special issue for crutches is stability. The user, feeling unstable or ready to fall, must be able to get his or her arms out of the device quickly.
4. *Aesthetics* –For a product this quality is of high importance. For an experimental setup, less so. However, the device must look accessible and not imposing.
5. *Experimental Context* –As this device is not an industry product, but an experimental apparatus, it must be easily adaptable to different experimental conditions. This means little time must be wasted on modification required to changing experimental conditions during an experimental session.

[3]For the experiments described in this thesis, we only refer to surface electromyography in which electrodes are placed over the muscle belly on the surface of the skin. While other more accurate forms of electromyography exist in which electrodes are implanted hypodermically, the implementation is too invasive for most dynamic studies.

Chapter 3

Experimental Methods

Clinical trials were held to evaluate the hypothesis that an elastic mechanism at the elbow can be used to optimize movement efficiency when crutching up stairs. First, the force-velocity characteristic of each subject's elbow joint was determined by collecting the maximum rotational velocities of the joint during isotonic flexion and extension. Subsequently, the subject's metabolic economy was measured while crutching up stairs. The metabolic experiments were repeated multiple times for each subject with elbow springs of different stiffness. This chapter provides a detailed description of the experimental methods including: (a) the design and implementation of the elbow-spring crutch, (b) the data collection equipment used to evaluate each subject, (c) information regarding subject recruitment and participation, and (d) data collection procedures for both the force-velocity experiments and the metabolic stair crutching experiments.

This study was approved by the MIT Committee on the Use of Humans as Experimental Subjects (COUHES).

3.1 Device Design

An elbow-spring crutch was designed in order to aid patients who require a chronic walking aid in stair climbing. The device is designed to allow the subject to climb stairs more efficiently (with a reduced metabolic rate) by placing a spring in parallel with the elbow joint.

3.1.1 Overview

The elbow-spring crutch, Figure 3-1, comprises three main elements: the cuffs, the spring, and the frame. During stair climbing, the device aids the user by applying an upward vertical force on the body. The cuffs mold around the biceps, as shown in Figure 3-2, allowing a load to be applied to the upper arm while limiting pressure applied to the adjacent muscle bellies. The springs extend from each side of the cuff to the handle circlet. Therefore, a flexing of the elbow as the hand is drawn towards the shoulder results in the compression of the spring. When the crutch is placed on

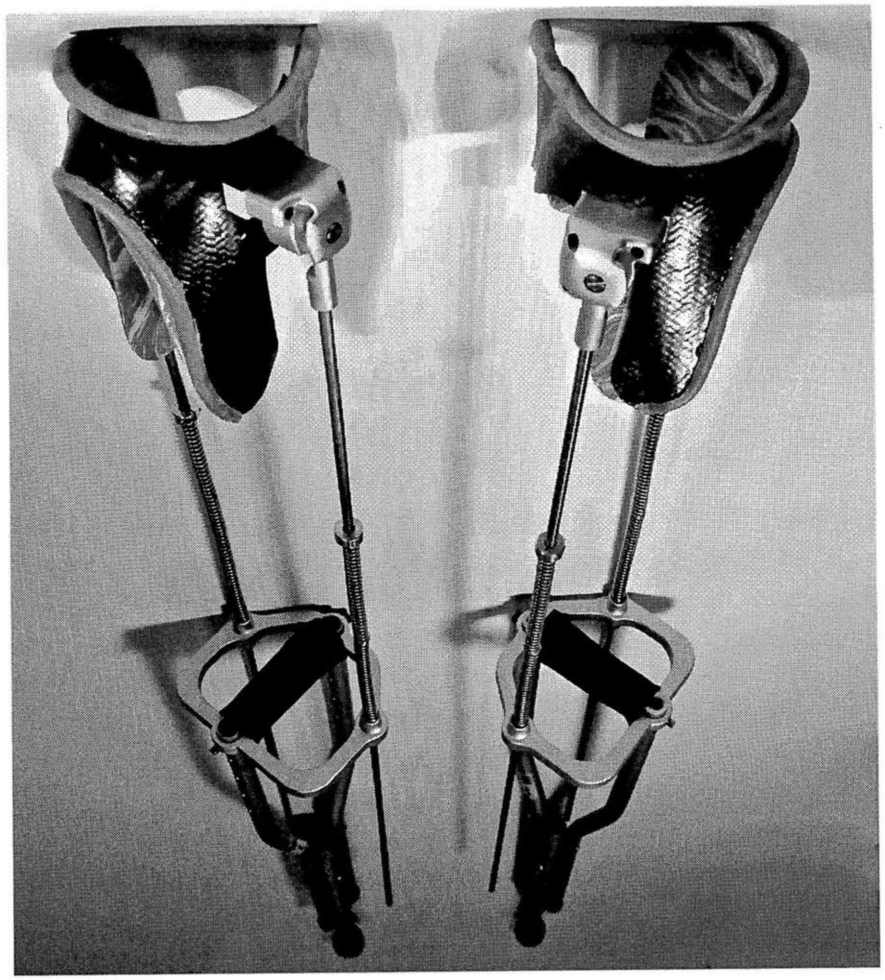


Figure 3-1: Elbow-spring crutches

the next stair, the energy stored in the compressed spring is released which in turn helps the muscle extensors straighten the elbow and push the body upward.

The overall weight of one elbow-spring crutch is 1.32 kg not including the slight difference in weight of the springs[1]. The vertical dimension can be customized to accommodate all subjects' heights via quick-release button connectors in the base.

3.1.2 Cuffs

The cuffs, Figure 3-3, are designed to apply an upward-vertical load on the upper arm without disturbing the contractions of the biceps or triceps. Additionally, the cuffs must not constrain the arm to the crutch in the case of a fall. Therefore, designs with straps and or slip-on features were discouraged. In consultations with a prosthetist, an organic wrapped shape with a rounded side-support was found to fit these functional requirements. To manufacture the cuffs, a negative plaster cast mold was made from

[1]As compared to a normal Lofstrand crutches which weight around 1.00kg.

Figure 3-2: Subject crutching up stairs with instrumentation gear. Electrodes locations are circled.

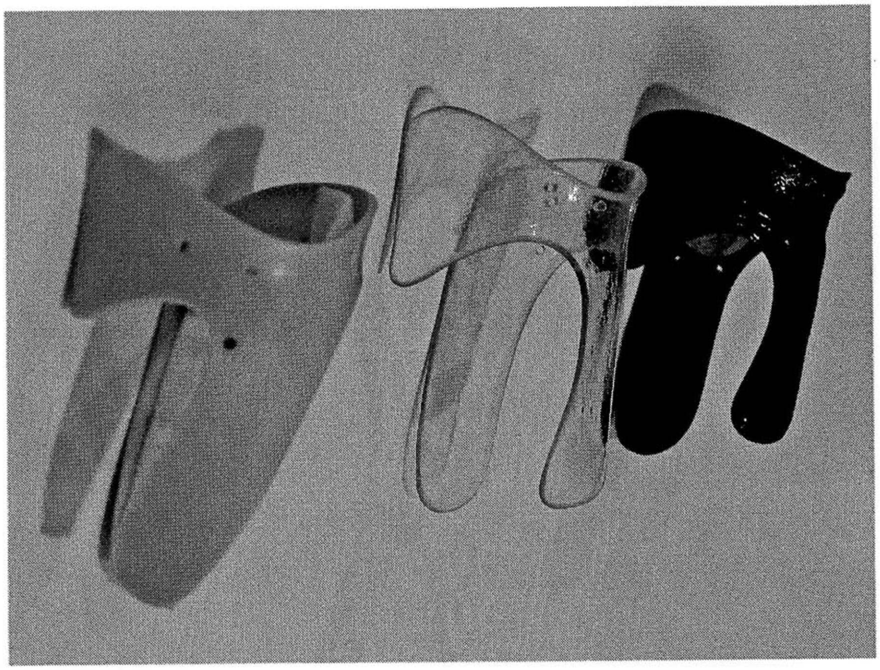


Figure 3-3: Crutch cuffs mold over the contours of the upper arm while avoiding the loading of muscles which are actively working during crutching. The above image, illustrates three iterations in cuff material selection. From left to right: co-poly cuff, plastic cuff, carbon fiber cuff.

a human arm. Subsequently, a positive mold was developed and served as the base for all future cuffs. The actual cuffs were manufactured via a vacuum-bagged carbon fiber wet layup on the positive mold.

Loading on soft tissues can be challenging. Vasculature and muscles need to be relatively unhindered in their natural operation. Large forces must be distributed over a wide area in order to reduce pressure points. Therefore, cushioning the hard carbon fiber shell is essential in order to provide sufficient user comfort. In the elbow-spring crutch, foam inserts on the cuff's inner face protects the subjects arm from pressure points, abrasion, and shear forces. Additionally, a rubber tubing trim along the cuff edges prevents pinch points along the cuff edges. During experimentation, extra foam pads were available to compensate for discomfort as needed.

3.1.3 Springs

In the elbow-spring crutch, the springs cannot impede arm motion and must be able easily switchable to change stiffness. While initial designs employed backward bending stacked leaf springs which were highly stable in the rotational direction, the geometry impeded natural elbow movement at full flexion. Furthermore, leaf springs are by default highly nonlinear which is hard to analyze in the experimental context. In contrast, linear springs can be easily positioned out of the way of the elbow and make discussion of forces more straightforward in an experimental setting[2].

A variety of springs were purchased from Century Spring[1], and combined in series, in order to achieve the right stiffness and throw length for each experimental condition. Table B.1 displays a list of all spring characteristics including spring stiffness, material, and quantities used in the current study. In the case of multiple stacked springs, the effective spring stiffness can be found via the following general constitutive equation derived from Hooke's Law:

$$K_{eff} = \frac{1}{\sum_{i=1}^{N} \frac{1}{K_i}} \tag{3.1}$$

where K_i refers to the individual spring stiffnesses and K_{eff} is the effective spring stiffness.

The table also lists the maximum force that the spring would apply in the course of normal operation (over the height of a normal stair).

3.1.4 Frame

The crutch frame supports the spring action about the elbow and supports the subject's body weight. This includes the crutch legs, handles, spring shafts, and spring mounts. The crutch leg is a base modified from an underarm crutch. The two-pronged base gives the subject extra support when applying additional spring loads through the floor. Telescoping tubes with quick-release buttons allow for quick height changes. On top of the crutch leg sits the handle circlet that supports the handle and four linear bearing mounts. The linear bearings (2 on top, 2 on bottom) align two shafts that stretch from the handle circlet to pin joints on either side of the cuff. The shafts are made of aluminum. The pin joints allow sagittal plane arm rotation from 37° to 157° as measured from the horizontal vector extending behind the body. Abduction and adduction of the arm is not inhibited in any way. During the initial design iterations, one possible option provided an extra degree of freedom at the wrist joint. However, this idea was quickly discarded as subjects reported that this decreased stability at a primary loading point which is highly undesirable.

3.2 Data Collection Tools

A variety of methods and sensors were used to evaluate the state of each subject. Metabolic, kinematic, force, video, and electromyographic data were collected at the instrumented Motion Analysis Laboratory (Holodeck) in the Computer Science and Artificial Intelligence Lab (CSAIL) at MIT.

[2]Although it should be noted that there is nothing to suggest that a nonlinear spring might be more or less optimal in this case.

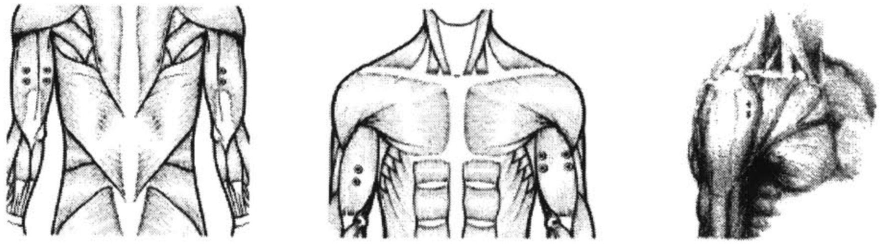


Figure 3-4: Electrode placements for upper arm EMG readings. A: Triceps bracii, B: Biceps bracii, C: Anterior deltoid. [8]

3.2.1 Electromyography

Surface electromyography (EMG) readings were taken from the biceps bracii, triceps bracii, and anterior deltoid muscles. Muscles from one randomly chosen arm were recorded synchronously at sampling rate of 1080Hz. A 16 channel EMG system and MA-411 preamplifiers (20X gain) from Motion Lab Systems were used for surface EMG measurements. The EMG data collection was done using disposable, pre-gelled surface bipolar electrodes with 20 mm center-to-center spacing (Electrode Store Model BS-24SAF, part # DDN-20). Electrodes were places on the center of the relevant muscle bellies in accordance with [8], Figure 3-4.

3.2.2 Motion Capture

Kinematic data for the experiments was recorded with a 16-camera motion capture system VICON 810i (Oxford Metrics @, Oxford, UK) at a frequency of 120Hz. The raw data was processed with VICON Work Station@ software. Reflective markers were mounted at 12 locations on the upper torso and limbs including the ulna and radius, medial and lateral elbow, shoulders, head and bicep. The spatial system's coordinate spacial resolution is approximately 2 mm. Post-experimental processing of marker position was completed in mathematical software MATLAB (MathWorks, Natick, MA) to compute marker velocities and accelerations.

3.2.3 Metabolic Rate

The rates of oxygen consumption and carbon dioxide production during each trial were measured using a portable metabolic analysis system (Cosmed K4b2, IT). Before each session, the gas analyzer was calibrated using reference gases and the flow-rate transducer was calibrated using a 3-liter syringe. Subjects completed one basal metabolic measurement (recording data while sitting with no activity) before each crutching session in order to better illustrate trial-to-trial variability in metabolic rate. In addition, subjects completed sessions at the same time of day and were requested to eat large meals at least 3 hours beforehand to reduce day-to-day variability in metabolic rate.

3.2.4 Force Transducer & Force Plates

During experimental trials an AMTI@ force platform system was used to synchronize the motion capture and force plate data. The force platform data was recorded at a sampling rate of 960HZ at an absolute precision of -0.1 N for vertical ground reaction force and -2mm for the center of pressure location with respect to the platform's center point. A stand-alone force transducer (Futek LRF350, 200lb Low Profile Tension & Compression unit) placed in line between the handle and the steel cable was employed to measure force at the human-machine interface. The analog force transducer data was recorded at 1000Hz and converted to a digital signal via a custom C# program.

3.3 Experimental Subjects

Four healthy adults participated in the study. The range of anatomical dimensions represented in the study are detailed in Table 3.1. For each subject, one arm was randomly chosen to be instrumented for data collection. Subject 2 was a trans-tibial bilateral amputee. None of the subjects had known elbow or shoulder disfunction or were chronic crutch users. Informed consent was obtained from all subjects.

3.3.1 Human Subject Use Approval

The experiments in this investigation were approved by the MIT Committee on the Use of Humans as Experimental Subjects (COUHES). Participation was strictly voluntary and subjects could withdraw from the study at any time. All subjects were asked to sign forms stating their informed consent before any experimentation took place.

3.4 Torque-Angular Velocity Characterization

3.4.1 Overview

The purpose of these experiments is to characterize the torque-angular velocity characteristic of the subject's elbow joint. Maximal angular velocity of the elbow was measured during isotonic flexions and extensions of the forearm during a crutching-like motion.

ject .e	Gender	Age	Body Weight	Forearm Length	Upper Arm Length	Wrist Circumference	Elbow Circumference	Shoulder Circumference	Tota Leng
	M	47	$60.0kg$	$0.27m$	$0.36m$	$0.15m$	$0.22m$	$0.29m$	0.98
	M	28	$73.0kg$	$0.30m$	$0.36m$	$0.17m$	$0.25m$	$0.34m$	1.15
	F	20	$57.0kg$	$0.24m$	$0.31m$	$0.15m$	$0.24m$	$0.30m$	0.92
	M	26	$63.5kg$	$0.26m$	$0.30m$	$0.165m$	$0.28m$	$0.31m$	1.02

(a) Patient Measured Anatomical Data

ject .e	Upper Arm		Forearm	
	Mass	Moment of Inertia	Mass	Moment of Inertia
	$1.87kg$	$0.0206m^2/kg$	$0.744kg$	$0.0045m^2/kg$
	$2.51kg$	$0.0279m^2/kg$	$1.07kg$	$0.0080m^2/kg$
	$1.78kg$	$0.0144m^2/kg$	$0.723kg$	$0.0033m^2/kg$
	$2.084kg$	$0.0167m^2/kg$	$1.05kg$	$0.0059m^2/kg$

(b) Patient Calculated Anatomical Data

3.1: Constants related to the patient's physical data. Data measured from the subject is listed in (a) and data calcu measured constants is listed in (b). Note: The moment of inertia is that of the body segment about it's biological rot

3.4.2 Data Collection Procedures

In this experiment the properties of the individuals' elbow joints were characterized by collecting force and kinematic data. These data are later combined with physical parameters of the subject taken using a biomechanical data collection form during the first session including: subject height, weight, limb length (upper-arm/forearm), and upper-arm/forearm circumference, Table 3.1.

3.4.3 Temporal Data Syncing

Simultaneously, ground reaction forces and kinematic data were recorded with the force plates and 3D motion capture system. This information was recorded by a 52i specific personal computer (Validation PC). The signals of the different sensory sources were synchronized at a base frequency of 120 Hz. In order to allow for time-syncing between the force transducer signal and the force plate system, before each trial the force transducer was placed on one of the force plates and tapped sharply with a mallet. The resulting impulses in both instrument signals were temporally synchronized in post-processing, Figure 3-5.

3.4.4 Experimental Apparatus

A commercial cable crossover machine (Body-Solid, Model PCCO90X) was modified to provide constant forces in the +/- vertical direction, Figure 3-6. A combination of high resin pulleys and steel cables transmit force between the handle and a sliding carriage on nylon bushings on the side of the device. When a force is applied to the handle, the carriage moves upwards stretching the attached lengths of latex tubing.[3]The lengths of latex tubing (VWR International, 1/4 (ID)x 5/8(OD)) were pre-stretched to approximately twice their resting length to provide a constant force on the carriage. The rigging of the steel cables is such that there is a 2x reduction in force from the input force on the carriage to the output at the location of the handle. A force transducer was attached in series with the cable at the level of the handle in order to provide a direct reading of the force applied at the end of the forearm.

3.4.5 Experimental Procedure

Subsequently to obtaining informed consent, subjects were asked to sit in a chair aligned with the handle of the cable crossover machine. After grabbing hold of the cable-crossover machine handle, each subject was instructed to flex or extend their elbow as fast as possible against the vertical force applied. The force was varied at approximately $10 - lb$ intervals from $F = 20lbs$ to $F = F_{max}$ where F_{max} is the maximum force a subject can pull when the velocity of the hand is zero (v = 0). Subjects were also asked to perform the motion in a zero force condition (without

[3]Here, latex springs replace traditional weights in order to reduce inertial effects from the acceleration of weight mass.

gripping the handle). Using these increments, two different types of loading conditions were imposed:

Flexion Loading Subject asked to lift arm (below shoulder) against constant force downwards

Extension Loading Subject asked to press down arm (below shoulder) against constant force upward

The flexion/extension motion imitates an actual crutching motion while climbing stairs, Figure 3-7. During the motion the arm was kept at a constant distance medial-laterally from the body center. Subjects were instructed to keep their arms adducted to their sides and their grip oriented anterior-posterior and horizontal with the floor (close as possible to sagittal plane motion). Extension and flexion occurred between 50° and 180°. If needed, mild restraints were used such that participants performed identical muscle movements.

Subjects were asked to complete the proper motion three times in a row with a break of at least two seconds in between each measured motion. The end of a trial was determined by the subject completing the proper motion set or three seconds passing, whichever came first. As an additional verification, isometric torque measurements for flexion and extension were taken separately by anchoring the force sensor in series with a length of steel cable and the same handle to the floor (or ceiling) and requesting that the subject resist as hard as possible against it. Isometric force measurements were taken at an elbow angle of 60°. A two minute rest period occurred between each trial and a 15 minute break between each set of experiments. All trials in this set were executed on the same day for an individual whenever possible. All of the force-velocity trials totaled around 4 hours per subject.

3.5 Stair Crutching Experiments

3.5.1 Overview

The purpose of these trials is to show a maximal metabolic advantage while crutching up stairs by tuning an elastic mechanism that spans the elbow joint.

3.5.2 Data Collection Procedures

Electromyographic, and metabolic data are collected throughout each trial. Subjects are also filmed via a digital video camera.

3.5.3 Experimental Procedure

Subjects were allowed to familiarize themselves with the equipment in previous sessions. Before each trial, subjects were instrumented with surface electrodes on the biceps, triceps, and anterior deltoid of the chosen arm. In order to calibrate the EMG signals in post-processing, 5s isometric maximal voluntary contractions (MVCs) were recorded for each muscle. The subjects arm was resisted by the investigator as detailed in [8] while subjects were encouraged verbally to give their maximal effort. Subsequently, subjects were fit with a mask and backpack from the Cosmed K4b2 metabolic analysis system. Masks were sized to the subject and lined with Vaseline to improve the seal where necessary. A basal metabolic test was run before each experimental session. During this test, the subject sits motionless for 7 minutes breathing into the mask with the Cosmed system recording.

After the basal metabolic measurements, subjects were asked to crutch nonstop up stairs to the beat of a metronome (cadence: 45 BPM), using the pair of modified Lofstrand crutches. While ambulating up a continuous staircase such as an escalator or a StairMaster® would have been ideal for nonstop crutching, realistic space requirements, both regarding the facility and the ergonomic use of crutches, prevented the application of these options. Thus, a set of two stairs were used and subjects alternated between crutching up a step and stepping down to the previous stair (without the use of crutches) while keeping a steady rhythm, Figure 3-8. The subjects executed each of the following motions in order executing exactly one motion per beat:

1. From dual stance[4], lift and place crutches onto the next stair.
2. Once positioned, press up with arms to lift the body until dual stance on the next stair.
3. Step down with both feet.

Subjects were instructed to keep arms adducted to the side of the body and to lift their elbows without raising the shoulder. Furthermore, as compensatory power plantar flexion (push-off) at the ankle could potentially confound trial results subjects were required to wear a semi-solid ankle foot orthosis (drop foot brace), Figure 3-9. The brace fits comfortably in a normal shoe and prevents the ankle from adding power to the crutching gait. Trials lasted for 7 minutes each. Subjects could request to stop a trial at any point. The subjects performed trials under the followings set of conditions:

Control The subject performs the experiment with the modified Canadian crutches (without springs).

Spring Tests The subject performs the experiment with the modified Canadian crutches (with springs). Three different spring conditions were chosen for each subject: the projected ideal spring stiffness (as calculated in Section 4.3), a spring stiffness greater than the ideal, and a spring stiffness less than the ideal.[5]

Each trial was repeated two or three times to guarantee accurate data. The order with which the spring trials were executed was randomized to rule out any

[4]In this case, dual stance refers to both crutches and biological legs on the ground.

sequential effects. The only exception was the highest spring stiffness tested which cause subjects some discomfort and therefore was left until last. A complete set of conditions for a particular subject were administered on the same day with a break of at least three days between each experimental session. Each session lasted around 3 hours. A sample experimental session schedule for a subject is shown below:

Sample Metabolic Experimental Session Schedule

Condition A

Basal Metabolic Test Rate 1

Crutching Test Spring, K = 0.25

Condition B

Basal Metabolic Test Rate 2

Crutching Test No Spring, K = 0

Condition C

Basal Metabolic Test Rate 3

Crutching Test Spring, K = 0.15

Condition D

Basal Metabolic Test Rate 4

Crutching Test Spring, K = 0.4

NOTE: All springs are reported here with their dimensionless spring values where $K_{dim} = \frac{KXm}{W}$, where K is the spring stiffness in N/m, X_m is the stair height, and W is the subject's weight.

[5]Springs were chosen as close as possible to the stated values based on manufacturer availability.

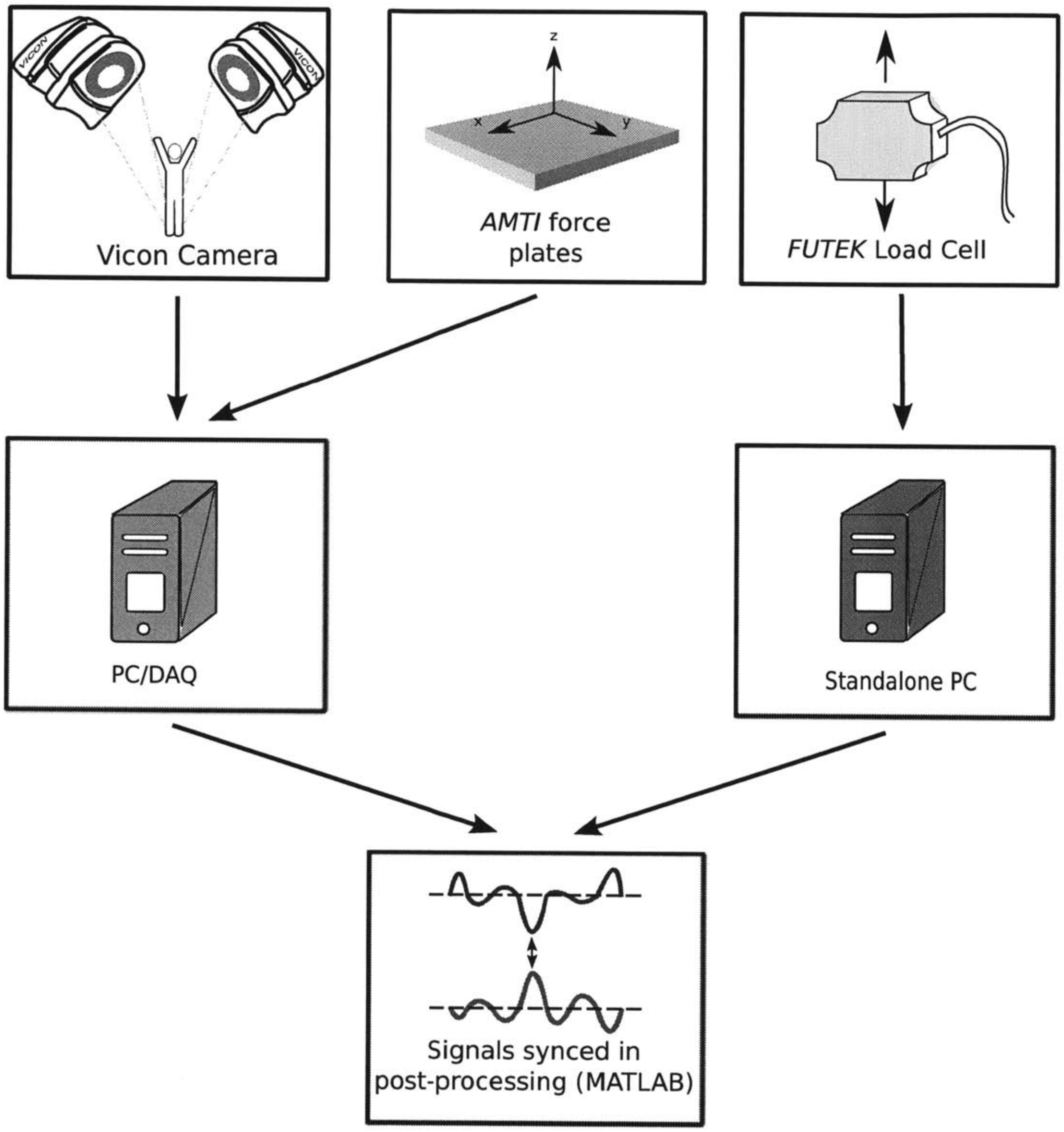


Figure 3-5: Data processing flowchart illustrating the flow of information from sensors to temporal syncing operation.

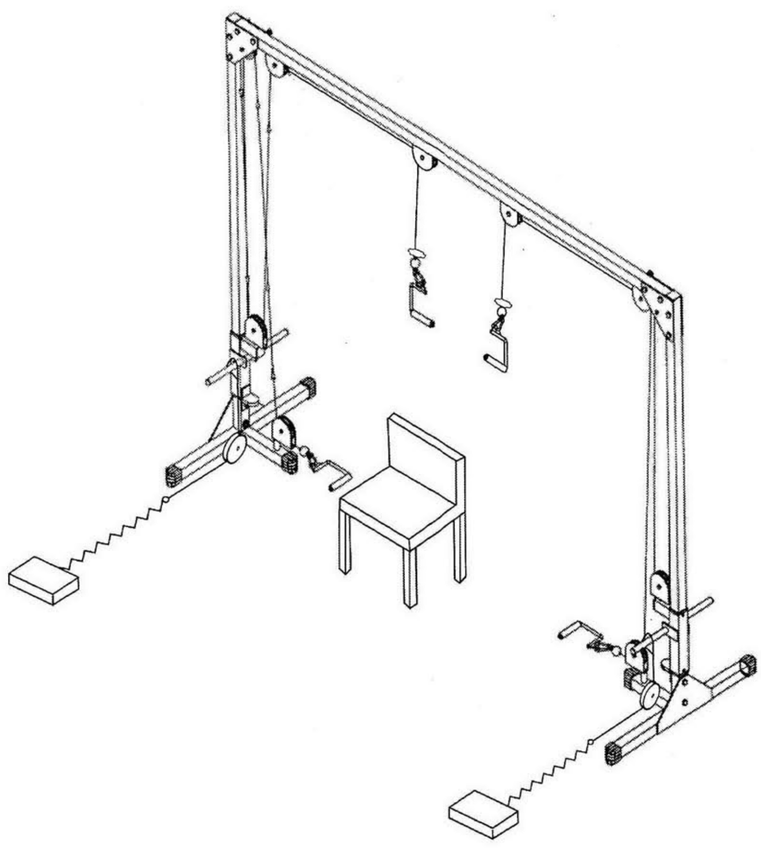


Figure 3-6: Modified cable-crossover machine. Pulleys direct forces vertically resisting arm motion with a series of latex springs. Handles can be hooked at the top or bottom locations for upward or downward vertical force resistance.

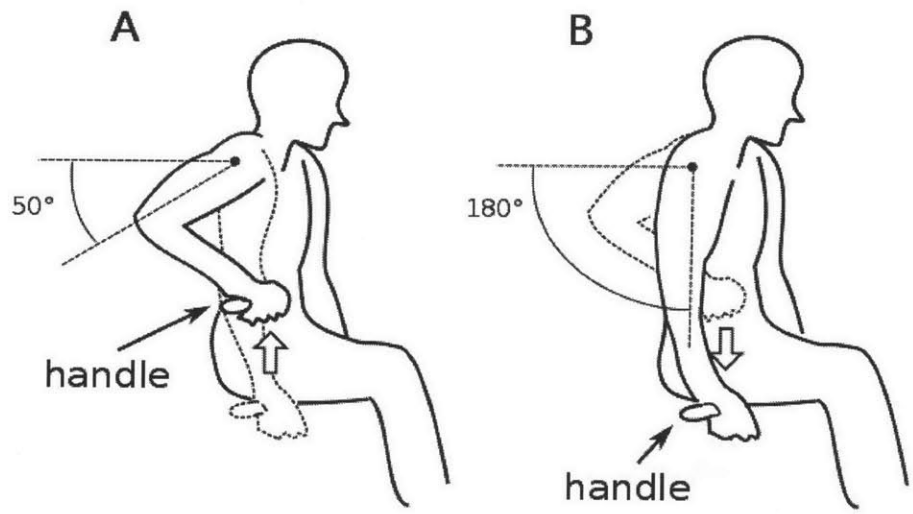


Figure 3-7: Crutching-like movement for muscle characterization. [A] Shows the biceps bracii characterization movement, [B] shows the triceps bracii characterization movement

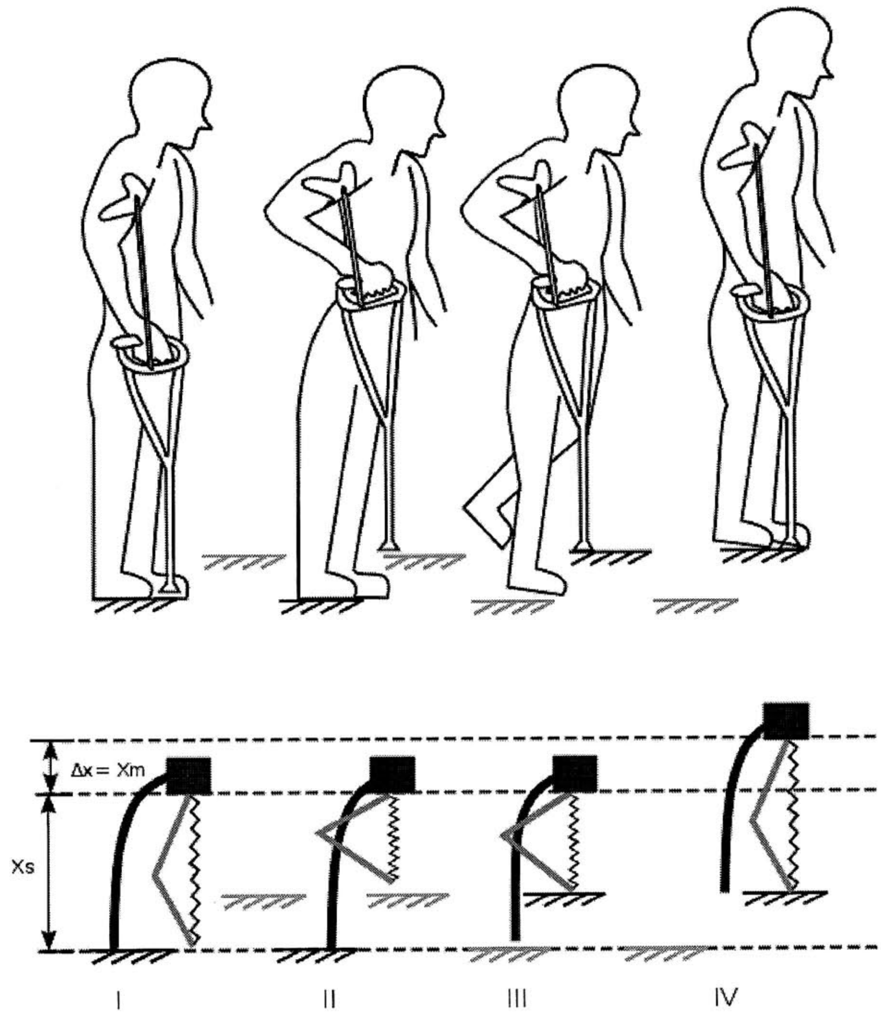


Figure 3-8: Weight transfer & spring compression in stair-climbing can be divided into four phases: I-IV. I: Crutch base is removed from the ground and the legs and body support the user. II: The elbow spring is compressed by flexing the elbow up to the next stair height. III: As the crutches are placed on the next stair, a shift of support occurs from the legs to the crutches. IV: The crutch spring expands helps the elbow flex, pushing up the user's body.

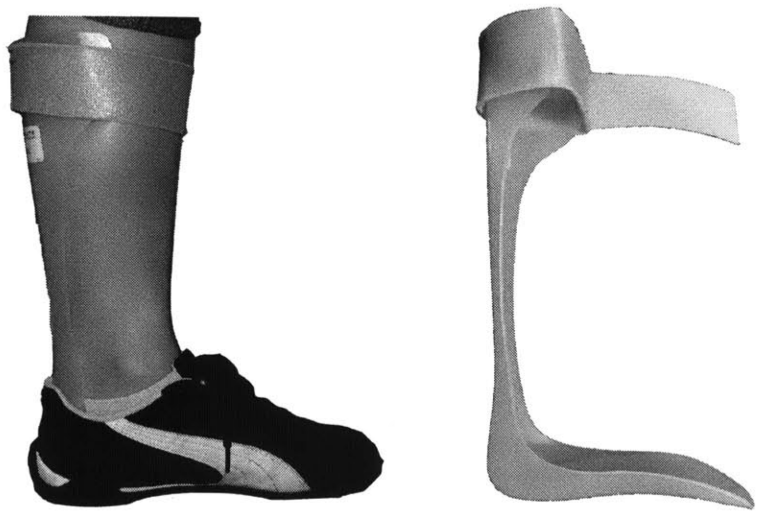


Figure 3-9: Ankle foot orthosis used to minimize power plantar flexion during push-off. [A] Subject wearing ankle foot orthosis. [B] Orthosis alone.

Chapter 4

Data Analysis

This chapter discusses the methods by which the raw data was analyzed. The chapter is divided into four sections: (a) Section 4.1 illustrates how one can calculate joint angles and torques from raw marker position data, including the set of assumptions employed to derive the equations of motion; (b) Section 4.2 discusses the fitting of the torque-angular velocity data to a hyperbolic curve; (c) Section 4.3 takes constants defined from this curve fit to predict an optimal spring stiffness; and (d) Section 4.4 describes the data processing of the experimental metabolic data to calculate an empirical optimal stiffness.

4.1 Arm Model

Sagittal plane arm movement anchored at the shoulder can be represented dynamically as an inverted double pendulum driven by joint torques at the shoulder and elbow. A simple model of a two-link manipulator with unique geometries for each link can be found in Figure 4-1 [29]. Both the upper arm and forearm links are approximated as elongated cone frustums as suggested in [6], Figure 4-2, with lengths, masses, and inertias l_1, m_1 , I_1 and l_2, m_2, I_2 respectively. Arm geometries are calculated using each subjects measured physical quantities, Table 3.1. This simple model is made more realistic by allowing the plane of the arm (the plane formed by the upper arm and forearm segments) to deviate slightly away from vertical during the course of a movement to compensate for any accidental arm abduction during experimentation, Figure 4-3. Given the described geometry, we can define the two angles of interest: the shoulder angle, θ_1, and the elbow angle, θ_2.

Since the joint of interest is a rotational interface, it makes more sense to characterize a subject's arm using a torque-angular velocity characteristic, rather than a translational force-velocity curve. Inverse kinematics and inverse dynamics analyses on the data were performed to obtain joint angle and torque time series. The joint angles can be derived from the raw marker positions as follows[1]:

$$
\begin{aligned}
\vec{n} &= \vec{l_1} \times \vec{l_2} & (4.1)\\
\hat{x}' &= \hat{z} \times \vec{n} & (4.2)\\
\hat{z}' &= \vec{n} \times \hat{x}' & (4.3)\\
\theta_1 &= tan_2^{-1}(\vec{l_1} \cdot -\hat{z}', \vec{l_1} \cdot -\hat{x}') & (4.4)\\
\theta_2 &= cos^{-1}(\frac{-\vec{l_1} \cdot \vec{l_2}}{|l_1||l_2|}) & (4.5)
\end{aligned}
$$

Joint velocities, $\dot{\theta}_1$ and $\dot{\theta}_2$, were then obtained by taking the central difference of the joint angle time-series. Using these angles and angular velocities, the corresponding muscle torques, τ_1 and τ_2, can be calculated by the following equations of motion represented in the manipulator form:

$$u = (H(q)\ddot{q} + C(q, \dot{q})\dot{q} + G(q) - F) \setminus B \tag{4.6}$$

Where,

$$q = \begin{bmatrix} \theta_1 \\ \theta_2 \end{bmatrix} \tag{4.7}$$

$$u = \begin{bmatrix} \tau_1 \\ \tau_2 \end{bmatrix} \tag{4.8}$$

The manipulator matrices can be written as follows:

$$H = \begin{bmatrix} H_{11} & H_{12} \\ H_{21} & H_{22} \end{bmatrix} \quad C = \begin{bmatrix} C_{11} & C_{12} \\ C_{21} & C_{22} \end{bmatrix} \quad G = \begin{bmatrix} G_1 \\ G_2 \end{bmatrix} \quad B = \begin{bmatrix} 1 & 0 \\ 0 & 1 \end{bmatrix} \quad F = \begin{bmatrix} F1 \\ F2 \end{bmatrix} \tag{4.9}$$

[1]Here, tan_2^{-1} refers to the four quadrant arctan which takes two parameters.

Where,

$$
\begin{aligned}
c_1 &= cos(\theta_1) && (4.10)\\
c_2 &= cos(\theta_2) && (4.11)\\
c_{1+2} &= cos(\theta_1+\theta_2) && (4.12)\\
s_1 &= sin(\theta_1) && (4.13)\\
s_2 &= sin(\theta_1) && (4.14)
\end{aligned}
$$

$$
\begin{aligned}
H_{11} &= m_1 r_{cm1}^2 + m_2(l_1^2 + r_{cm2}^2 + 2 l_1 r_{cm2} c_2) + I_1 + I_2 && (4.15)\\
H_{22} &= m_2 r_{cm2}^2 + I_2 && (4.16)\\
H_{12} &= m_2(r_{cm2}^2 + l_1 r_{cm2} c_2) + I_2 && (4.17)\\
H_{21} &= H_{12} && (4.18)
\end{aligned}
$$

$$
\begin{aligned}
C_{11} &= -2 m_2 l_1 r_{cm2} s_2 \dot{\theta}_2 && (4.19)\\
C_{22} &= 0 && (4.20)\\
C_{12} &= -m_2 l_1 r_{cm2} s_2 \dot{\theta}_2 && (4.21)\\
C_{21} &= m_2 l_1 r_{cm2} s_2 \dot{\theta}_1 && (4.22)
\end{aligned}
$$

$$
\begin{aligned}
G_1 &= (m_1 r_{cm1} + m_2 l_1) g c_1 + m_2 r_{cm2} g c_{1+2} && (4.23)\\
G_2 &= m_2 r_{cm2} g c_{1+2} && (4.24)
\end{aligned}
$$

$$
\begin{aligned}
F_1 &= f_0(cos(\theta_1) + cos(\theta_1+\theta_2))\theta_1 && (4.25)\\
F_2 &= f_0 cos(\theta_1+\theta_2)\theta_2 && (4.26)
\end{aligned}
$$

NOTES:

1. The values r_{cm1} and r_{cm2} refer to the distance from the parent joint to the center of mass of the segment.

2. f_0 corresponds to the external vertical force applied to the subject at the handle-wrist interface.

4.2 Torque-Angular Velocity Data Processing Methods

As described in Section 3.4.5, a subjects's elbow joint characterization is based on three joint extensions and three joint flexions under a series of isotonic loads. This

[1]Here, the term "vertical" should be interpreted as the vector orthogonal to the horizontal vector in the plane of the arm. With the given experimental constraints, $\hat{(z)}'$ was within 10o degrees of vertical for most tests.

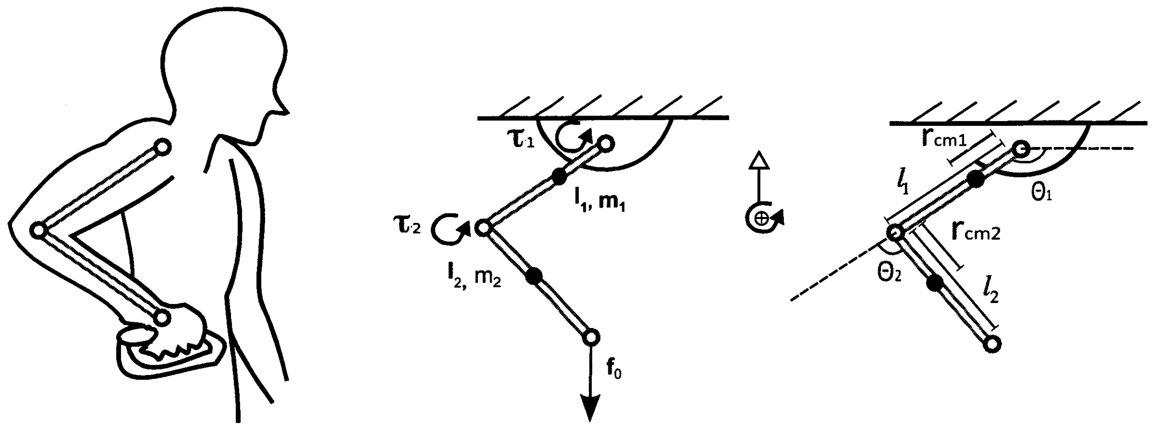


Figure 4-1: Diagram of arm model

data collection procedure produces two asynchronous signals from the load cell and the Vicon system. The signals are synchronized by time-shifting such that the impulse features, described in Section 3.4.3, align to within 0.01 seconds accuracy.

As one might expect, each of the three flexion and extension movements produces a peak in the angular velocity signal. The angular velocity value at a particular force is calculated by taking the average of 30 data points[2] surrounding each curve peak. This magnitude is subsequently averaged again over the 3 peaks in order to reduce measurement error . Subsequently, the torque amplitudes are calculated by averaging the torque signal values from the corresponding time points.

By plotting the measured elbow angular velocity, $\dot{\theta}_2$, against the calculated elbow torque, τ_2, we expect the data to follow a rectangular hyperbola as described in Section 2.2.2. A three dimensional regression analysis (variables: τ_2, θ_2, and $\tau_2 * \theta_2$) is used to find the best-fit curve[36]. From the best fit curve parameters, we can approximate F_0, V_m, and a, the negation of the location of curve's vertical asymptote with which to calculate the optimal spring stiffness as described in Section 4.3.

4.3 Predicting an Optimal Spring Stiffness

Complex movements often make use of muscles in tandem rather than one or two muscles operating individually. When considering upper arm movements like stair-crutching, defining an "effective" muscle can be a useful way to analyze the muscle system acting as a whole. Within this paradigm, force-velocity data from the effective muscle can be used to develop an energetics model that predicts the spring stiffness which produces maximally efficient movements. Herr [12] uses such a model to great effect for highly fatiguing dynamic motions. In his study on endurance amplification, he predicts the stiffness which corresponds to optimally efficiency movements and

[2]An arbitrary number that is small enough to not exceed the width of the smallest peaks while large enough to encompass any reasonable variation in magnitude.

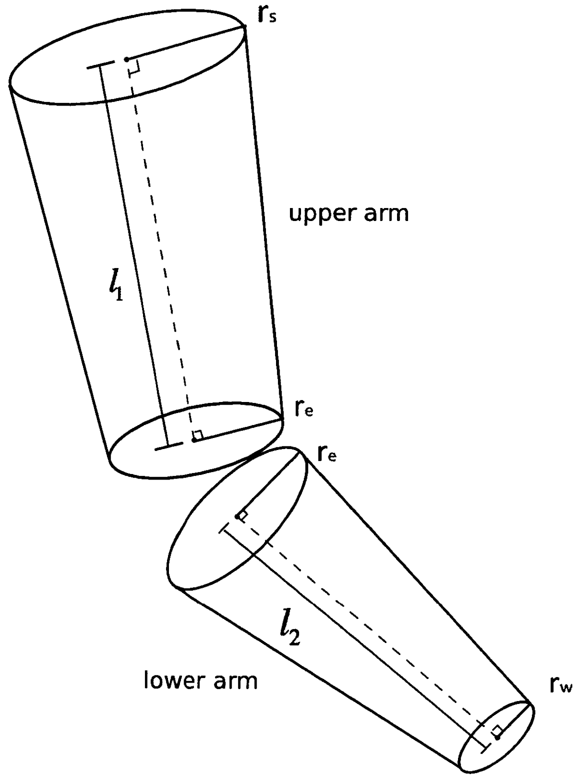


Figure 4-2: Arm segments are approximated as cone frustums for the purpose of calculating the masses and inertias used in the dynamics model

uses that to anticipate the empirical value found to maximize endurance (cycles until exhaustion) for the same motions. This study is fundamentally different as, here, we compare energetics model results to an energetic measurement ($\dot{V}_{O2}$ data) over the course of the experiment rather than looking at fatigue-based endurance. However, we expect similar results.

For the purpose of these experiments we define efficiency as:

$$\varepsilon = \frac{\omega_{arm}}{E_{met}} \tag{4.27}$$

where, ω_{arm} is defined as the muscle work required to extend and flex the arm during one crutching cycle and E_{met} refers to the metabolic energy liberated per cycle. As each arm compresses the elbow spring and, then, lifts half the subject's body weight during and once each cycle ω_{arm} can be approximated by:

$$\omega_{arm} = \frac{1}{2}WX_m + \frac{1}{2}KX_m^2 \tag{4.28}$$

In order to predict an optimal spring spring stiffness which maximizes the energy efficiency in stair-crutching, a distribution-moment model of skeletal muscle [20] is used to calculate a dimensionless metabolic rate, Ω, for constant velocity muscle contractions. Using empirical methods described in [20], a curve plotting this metabolic rate vs. normalized muscle velocity can be determined and is reproduced in Figure 4-4. Ω can also be defined as the metabolic rate $\dot{E}$ normalized by the product of the isometric muscle force, F_0, and the maximum muscle contraction velocity, V_m:

$$\Omega = \frac{\dot{E}}{(F_0 V_m)} \tag{4.29}$$

Using Ω, the experimentally determined constants F_0 and V_m, and the distance of spring compression (in this case, the stair height, X_m) one can express E_{met}, in terms of the muscle shortening velocity, V:

$$E_{met} = \int_0^{X_m} \{F_0\,(\frac{V_m}{V})\,\Omega\}_{ext}\,\mathrm{d}X + \int_0^{X_m} \{F_0\,(\frac{V_m}{V})\,\Omega\}_{flex}\,\mathrm{d}X \tag{4.30}$$

In (4.30), the two terms correspond to the metabolic energy consumed during arm extension and arm flexion respectively. In order to find the optimal spring stiffness, Eq. 4.30, must be expressed in terms of dimensionless elbow spring stiffness, K_{norm}.

Using the force-velocity characteristic of the arm collected as described in Section 3.4, the composite force-velocity properties of the elbow effective muscle are found to be generally hyperbolic which mirrors known properties of isolated skeletal muscle. Using the notation from Eq. 2.2, the normalized effective arm muscle velocity can be represented as:

$$V' = \frac{V}{V_m} = \frac{M_1 - (\frac{F}{W})}{M_1 + M_2(\frac{F}{W})} \tag{4.31}$$

where $M_1 = \frac{F_0}{W}$, $M_2 = \frac{F_0}{a}$.

Finally, one can derive relationship between effective muscle force and dimensionless arm spring stiffness. During the first half of the cycle when the subject lifts flexes the elbow joint, the force exerted on the forearm (normalized by body weight) is that required to compress the elbow spring a distance X_m:

$$F' = \frac{F_e}{W} = K(\frac{X}{X_m}) \tag{4.32}$$

During the second half of the cycle, the subject extends both arms, pushing the crutch legs against the ground and lifting the body up to the next stair. The force (normalized by body weight) is equal to:

$$F' = \frac{F_f}{W} = \frac{1}{2} - K(\frac{X}{X_m}) \tag{4.33}$$

By combining equations 4.30 through 4.33 and integrating, one can graph, the metabolic efficiency vs. the elbow spring stiffness, Figure 5-2.

4.4 Metabolic Data Processing Methods

4.4.1 Net Metabolic Power

Normalized metabolic power was computed from the collected data via the following energy expenditure equation [7]:

$$P = \frac{16.58(\dot{V}_{O_2avg}) + 4.51(\dot{V}_{CO_2avg})}{60W} \tag{4.34}$$

where P refers to the total bodily power expenditure in W/kg, W corresponds to the subject's mass in kg and $\dot{V}_{O_2avg}$ and $\dot{V}_{CO_2avg}$ are the average oxygen gas exchange rate (mL/min) and the average carbon dioxide gas exchange rate (mL/min), respectively.

The values $\dot{V}_{O_2avg}$ and $\dot{V}_{CO_2avg}$ for a particular trial were calculated over a particular time interval. Within that range, both breath holds (where no data was recorded by the COSMED system) and zero readings (read errors corresponding to leaks in the subject's face mask) were ignored. The proper time interval was chosen using the following criteria:

1. The interval must start after the signal has achieved steady state (after the 2 min mark for all trials)

2. The time interval must not extend into the last 30 seconds of the trial

3. The time interval must be as long as possible with a minimum length of 2 min 30 sec

Corresponding basal power measurements were subtracted from their subsequent crutching trial measurements (gross metabolic power) yielding the net power for the crutching activity.

$$P_{net} = P_{crutching} - P_{basal} \tag{4.35}$$

Combining P_{net} with experimental constants we can calculate the dimensionless specific cost of transport or c_t which is normally described as the energy required to move a unit weight a unit distance (in this case to lift 1kg up 1 step):

$$c_t = \frac{60(P_{net})(BPS)}{g(BPM)(X_m)} \tag{4.36}$$

where BPS refers to the cadence (the number of metronome beats per step climbed), BPM refers to the metronome timing in beats/min, X_m is the step height in m, and g is the gravitational constant.

4.4.2 Respiratory Exchange Rate (RER)

Concurrently, the respiratory exchange ratio (RER) was monitored for all trials. RER can be calculated as follows:

$$RER = \frac{\dot{V}_{CO_2avg}}{\dot{V}_{O_2avg}} \tag{4.37}$$

RER values for all experiments were less than 1 indicating that metabolic activity was primarily supplied by oxidative metabolism rather than anaerobic muscle activity.

4.5 EMG

While many mathematical approaches have been proposed for using EMG data to estimate muscle activation, the most method is based on the idea that the envelope of the demodulated EMG signal is an acceptable measure of muscle activity, as this processed signal is a slow process compared to the spiky neural excitation events captured in raw data[16]. The amplitude envelope is often extracted by applying a low-pass filter on the full wave- rectified EMG signal. Since we do not claim to use this signal for any higher purpose (such as control) and only look for a qualitative comparison of muscle activity, this simple method is more than sufficient.

In this study, the EMG signals were recorded from sites on the subject's biceps bracii, triceps bracii, and anterior deltoid. Initially, they were processed with a 2nd order Butterworth bandpass filter with cutoff frequencies at 10hz and 500hz (the frequency band of relevant biological activity). The signals were then zeroed, rectified, and normalized by their respective maximal voluntary contraction (MVC) values. The final product was smoothed with a moving average method with a frame window of 10% of the sampling frequency. The MVC values were obtained by integrating the root-mean-square (RMS) of the raw MVC signal during muscle contraction over a 5 second interval. A relative comparison of muscle activation can be obtained by averaging the area underneath 5 sample signal bursts from one muscle during a trial and comparing to other trials, Figure 5-6.

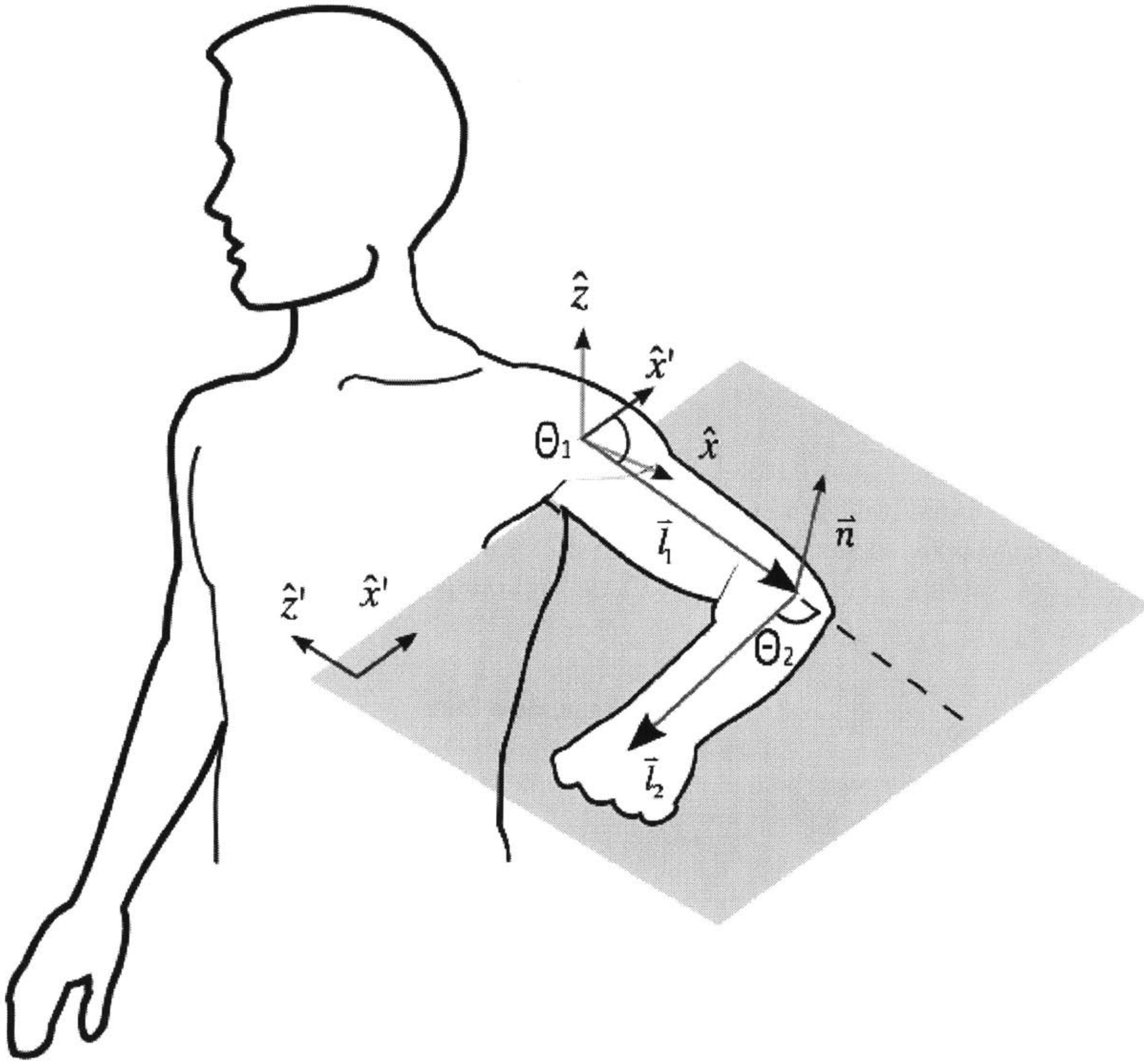


Figure 4-3: Vector geometry used in arm model. The vectors $\vec{l_1}$ and $\vec{l_2}$ are generated from motion capture data and lie along the upper and lower arm respectively. $\vec{n}$ is the normal vector of the plane formed by the two arm segments. The unit vectors $\hat{x}$ and $\hat{z}$ represent the horizontal and vertical directions in the lab-frame. $\hat{x}'$ and $\hat{z}'$ correspond to the vertical and horizontal vectors in the plane of the arm. From these vectors, we can define θ_1 as the angle between $\hat{x}'$ and $\vec{l_1}$ and θ_2 as the relative angle between forearm and the projected continuation of the upper arm.

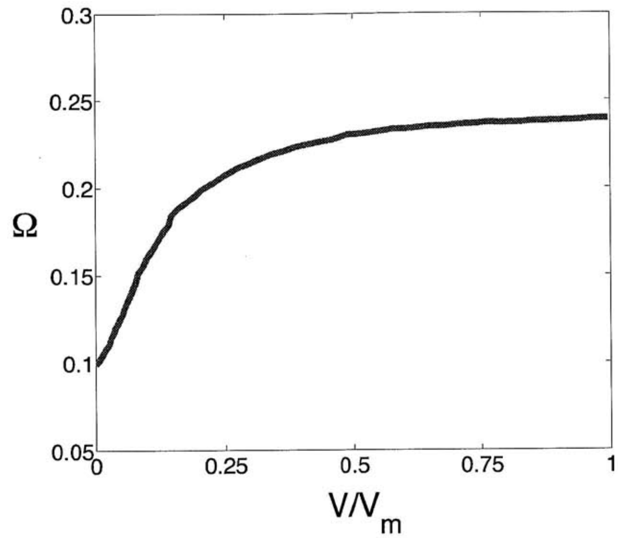


Figure 4-4: Dimensionless metabolic rate as a function of muscle contraction velocity. The velocity term is normalized by the maximum muscle contraction velocity (a muscle contracting against no load). Curve reproduced from [20]

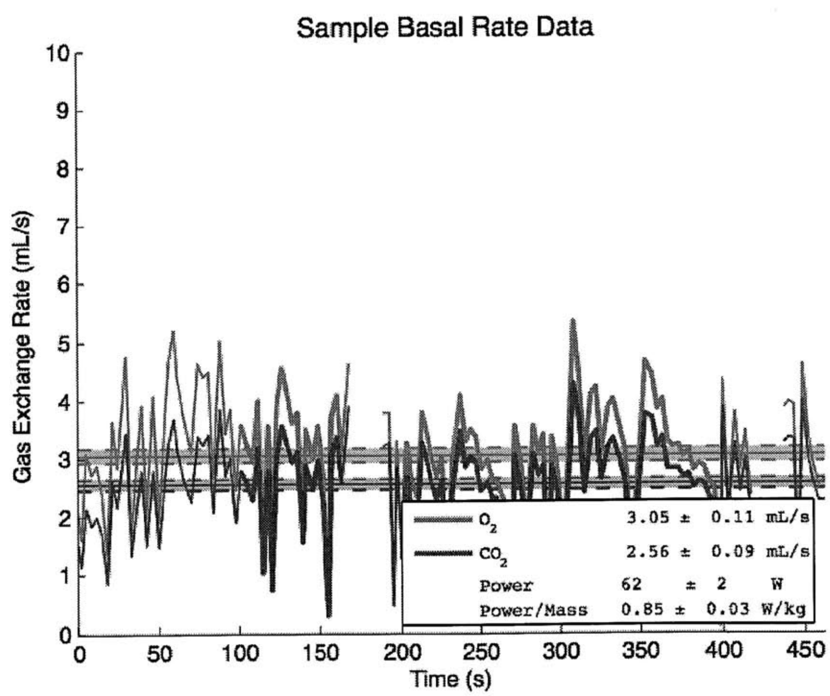


Figure 4-5: Sample basal metabolic rate data. Bold areas highlight the segment of the signal used for analysis.

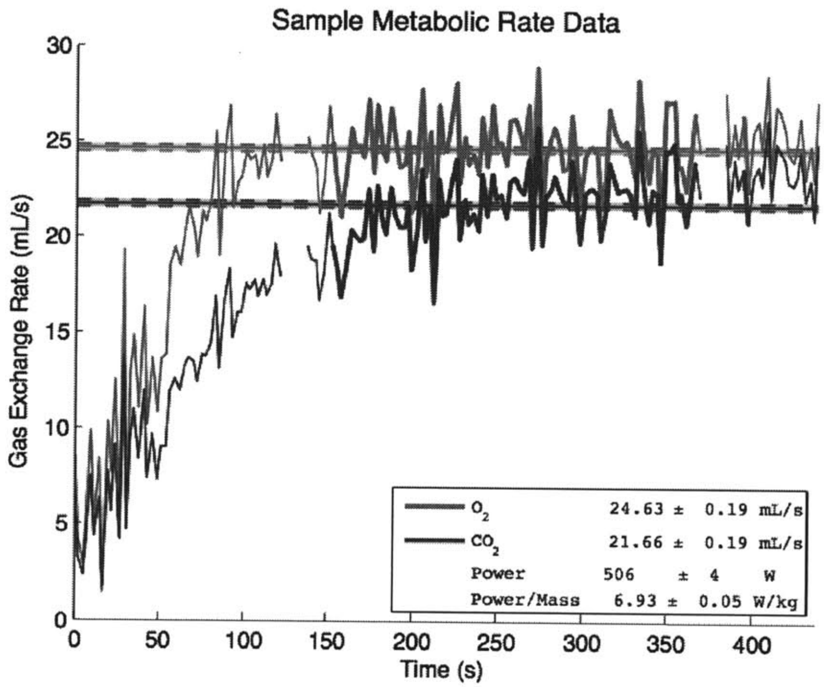


Figure 4-6: Sample metabolic rate data while crutching. Bold areas highlight the segment of the signal used for analysis.

Chapter 5

Results

This chapter exhibits the results from the two experiments described in Chapter 3. First, we present the torque-angular velocity characterizations for the flexor and extensor muscles. For each subject, we analyze the predicted optimal spring stiffness produced by the energetics model. Then, we compare the results of the model prediction with the empirical metabolic results (both the individual datasets and the combined dataset for all subjects). Finally, we present the results of the EMG measurements. Specifically we discuss the insight that electromyography provides with respect to the dynamic interactions at the muscular level. We end with a discussion of the results.

5.1 Torque-Angular Velocity Characterization Results

The best fit hyperbolic curves were calculated for each subject's torque-angular velocity data, Figure 5-1 & Appendix A. The data fit well to the model with $R^2 > 0.8$ in all cases. The constants τ_0 and a were derived from each subject's hyperbolic fit, Table 5.1. For all subjects, the isometric torque constant, τ_0 is greater for flexors than extensors which suggests that the biceps can ultimately produce larger forces in this bodily configuration. The constant a gives a sense of how fast the subject can "contract" the effective muscle in the other isotonic measurements relative to the isometric case.

The energetic model's maximum efficiency predictions are shown in Figure 5-2. For all subjects, the model exhibits a maximal[1] efficiency between 5-25%. Predicted optimal K_{norm} values range from 0.287-0.656 with a mean of 0.416 and a standard deviation of 0.165. Experiments revealed empirical optimal K-values on average 0.225 stiffness units lower than the model prediction.

[1]Note: Metabolic reduction is by definition inversely correlated with efficiency.

Subject Code	Muscle	τ_0	a
1	*biceps*	$143Nm$	33.9
	triceps	$128Nm$	48.6
2	*biceps*	$187Nm$	59.5
	triceps	$153Nm$	66.4
3	*biceps*	$141Nm$	47.5
	triceps	$103Nm$	20.4
4	*biceps*	$156Nm$	21.6
	triceps	$105Nm$	22.7

Table 5.1: Torque-Angular Velocity Characteristic Constants

5.2 Metabolic Results

The results of the metabolic tests are summarized in Tables 5.2 & 5.3 and Figures 5-3 & 5-4. All average RER values are below 1. Metabolic measurement error rates did not exceed 5%. While the magnitudes of the metabolic data are difficult to compare between individuals due to differences in metabolism and experimental conditions, one can compare the relative shape of the distributions. All subjects produced a minimum power value in the center of the distribution with the exception of Subject 4 who's metabolic rate barely dropped below nominal for any trial. This suggests that all spring stiffnesses conditions were too strong and thus Subject 3 did not enter into the space of K values which would produce a metabolic reduction. For the rest of the subjects, the minimum power value occurred at or near a mean K_{norm} of $\mu = 0.1904$ with a standard deviation of $\sigma = 0.0414$.

In general, after the experimental session, subjects reported that stair ascent became easier as the spring stiffness was increased, even as it felt harder to compress the spring in preparation.

5.2.1 Compiled Dataset

While it makes little sense to analyze curve fitting for an individual subject with few data points, this is not a concern for the compiled dataset. We suggest a quadratic as a reasonable curve fitting model as it is a function with the fewest parameters that can achieve a minimum. Here, we restrict the fit to points local to the measured minima, within a range of $+/-0.2K_{norm}$. From the fit curve, one can project an global optimal local minimum of $K_{opt} = 0.133$ and a global optimal metabolic reduction of $P_{opt} = 84\%$ of $P_{nominal}$ (at $K_{norm} = 0$).

5.2.2 Statistics

In order to test for significance in the combined dataset, the data was binned into sequential[2] overlapping bins of four samples each such that the first four data points

Subject Code	K_{norm}	Basal Power	Net Power	RER	c_t
1	0	1.45 W/kg	5.37 W/kg	0.884	14.4
	0.2270	**1.34 W/kg**	**4.98 W/kg**	**0.886**	**13.3**
	0.2812	1.38W/kg	5.56 W/kg	0.811	14.9
	0.6718	1.49W/kg	6.70 W/kg	0.789	17.9
2	0	0.91 W/kg	6.25 W/kg	0.892	16.7
	0.1865	**0.86 W/kg**	**4.42 W/kg**	**0.780**	**11.9**
	0.2323	1.13 W/kg	5.83 W/kg	0.884	15.6
	0.5521	1.01 W/kg	5.98 W/kg	0.865	16.0
3	0	0.84 W/kg	3.67 W/kg	0.917	9.84
	0.1338	**0.69 W/kg**	**3.37 W/kg**	**0.921**	**9.02**
	0.2389	0.66 W/kg	3.51 W/kg	0.983	9.40
	0.2962	0.78 W/kg	3.69 W/kg	0.984	9.88
4	0	0.75 W/kg	5.48 W/kg	0.967	14.7
	0.2144	0.82 W/kg	5.50 W/kg	0.891	14.7
	0.2659	**0.88 W/kg**	**5.42 W/kg**	**0.896**	**14.5**
	0.6347	0.74 W/kg	6.49 W/kg	0.836	17.4

Table 5.2: Metabolic data taken from the COSMED oxygen analysis system. Minimum net power measurement data is highlighted in bold.

Subject Code	K_{norm}	Measured Maximum Metabolic Reduction	Projected Maximum Metabolic Reduction
1	0.227	7.4%	8.8%
2	0.187	29.2%	12.4%
3	0.134	8.0%	15.3%
4	0.266	1%	2.0%
Total	**0.211**	**6.6%**	**15.2%**

Table 5.3: Actual and projected percent metabolic reduction due to the addition of springs. For each subject, metabolic reduction is calculated as a percentage relative to the control P_{net} measured at K = 0. Projected metabolic reduction is extrapolated using a quadratic fit to the measured data points.

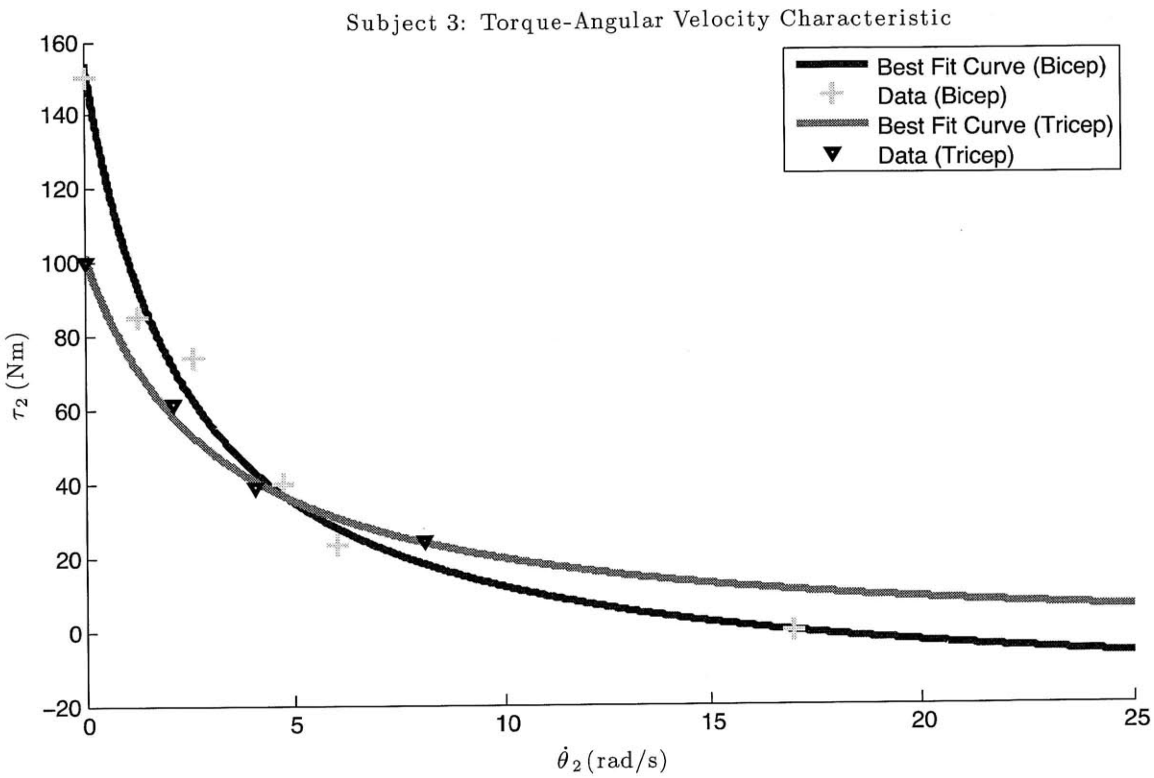


Figure 5-1: A sample torque-angular velocity characteristic for the effective elbow muscle. The extensor (triceps) is pictured in red and the flexor (biceps) is displayed in blue. The best-fit hyperbolic function is plotted over the raw data. The extensor and flexor curve fits had R^2 values of 0.84 and 0.97 respectively.

along the K_{norm} axis appear in the first bin, data points 2-4 appear in the second bin, 3-6 appear in the third bin, etc. Here, we make the approximation that the data in each bin is representative of a single gaussian distribution (an approximation that becomes more accurate with more data points along the K_{norm} axis). Then, the data in each bin was compared to the nominal with a paired t-test. The resulting p-values are plotted in Figure 5-5. One can see that the p-values dip toward the center of the distribution, arriving at a clear minimum which tests statistically significant. Certainly, one bin's significance does not prove significance for the effect over the entire study. However, while the currently available data is not by itself sufficient to demonstrate a significant deviation, the data shows a trend towards significance in the direction expected.

[2]Here, sequential refers to the location along the K_{norm} axis.

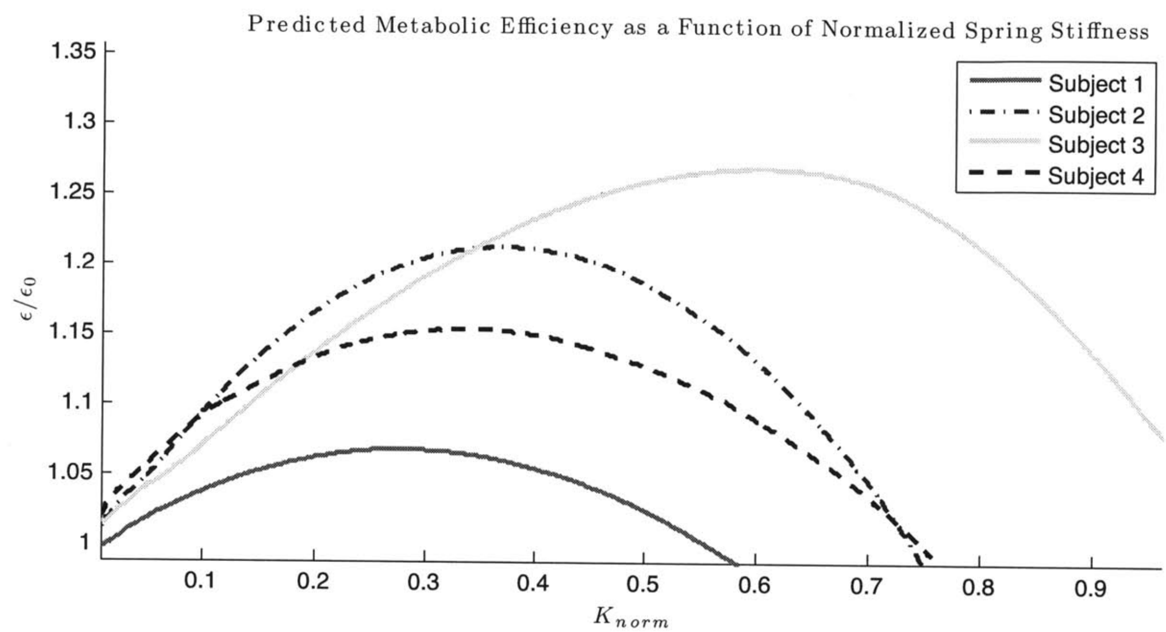


Figure 5-2: Predicted optimal spring stiffness as a function of normalized stiffness for all four subjects.

5.3 EMG Results

Electromyography looks at the neural control signal commanding individual muscle activity. EMG data was recorded for each subject during the full length of the trial. The signal exhibits a characteristic pattern over the course of a gait cycle. The first smaller peak represents the biceps activating as the elbow flexes and the crutch spring is compressed, Figure 5-6. The second larger peak marks the period where the triceps engage, extending the elbow and lifting the body up to the next stair. As shown in the following figures, there is very little co-contraction present in either movement. Uncontrollable variations in experimental conditions such as local skin conductance and individual muscle morphology, suggest that realistically these measurements cannot be compared among subjects. However, one can compare data between trials taken from the same subject under different stiffness conditions given the MVC normaliza-

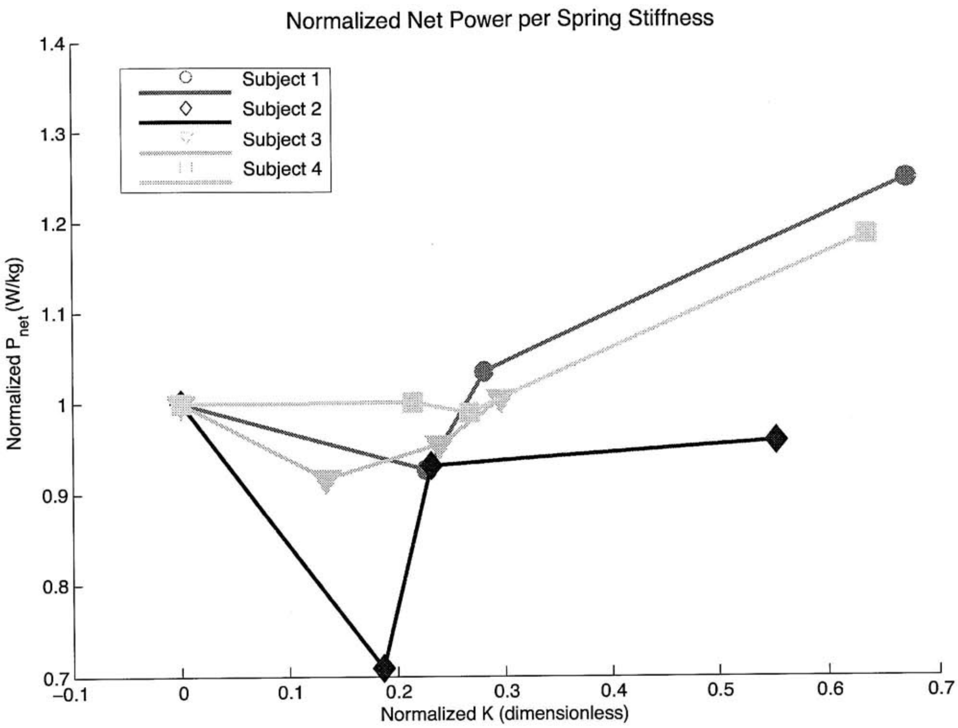


Figure 5-3: Metabolic rate results. Metabolic power consumption as a function of normalized spring stiffness. Power is normalized by the control, in this case the magnitude at $K_{norm} = 0$.

tion. A full set of MVC signals are displayed in Figure, 5-7. A sample set of muscle activation levels of one subject is listed in Table 5.4. Here, the activation magnitudes are listed in arbitrary units and should be considered qualitative rather than quantitative data. Even so, one can clearly mark a trend where the bicep activation increases monotonically and the triceps activation decreases monotonically as the elbow springs become stiffer, Figure 5-8 though 5-9. It is interesting to note that sum of the muscle activation magnitudes decrease by over 50% as stiffness varies from zero to K_{max}, however this cannot truly be verified without a more comprehensive dataset of multiple trials for an individual subject.

5.4 Discussion

5.4.1 Metabolic Reduction

The observed results show a trend towards metabolic reduction over a range of elbow spring stiffnesses. The empirical data illustrate a trend with a projected minimum metabolic reduction at $K_{opt} = 0.133$. More subjects would be needed to make the results significant. While all four subjects show some form of metabolic reduction, Subject 2 shows the largest response (70% reduction compared with the next lowest value at 8%). It is possible that this disparity results from the fact that Subject 2 could not use his ankles to plantar flex during push-off, minimizing any help provided by the lower leg. During experimentation, it was clear that the other subject's ankle-foot

Muscle	K_{norm}	Muscle Activation	Activation Change
Bicep	0	0.1912	0%
	0.214	1.6722	+8.74%
	0.266	4.9534	+25.9%
	0.635	5.8103	+30.4%
Tricep	0	1.5391	0%
	0.214	1.4045	−8.74%
	0.266	0.7964	−48.3%
	0.635	0.5586	−63.7%

Table 5.4: Comparison of muscle activation over different spring stiffness for primary arm muslces. NOTE: Activation Change is the percent change in muscle activation as compared to the control ($K_{norm} = 0$)

orthoses only minimized, but did not fully prevent, ankle plantar flexion. Parasitic energy loss at the ankle and the introduction of muscle groups unrelated to the elbow joint would significantly reduce a subject's potential for maximal metabolic reduction during push-off. In future tests, it would be suggested that subjects be asked to wear a completely rigid orthosis in order to prevent this effect.

5.4.2 Predicted vs. Measured Optimal

From [12], we would expect the model to accurately predict the location of the optimal stiffness value. However, the results illustrate a notable difference between the predicted and the empirical K_{opt}. As discussed in the previous section, unmodeled ankle torque could confound results by reducing the magnitude of the measured metabolic reduction. This effect would make the extension phase of the crutching action easier, reducing the spring stiffness at which one would measure an optimal reduction. Furthermore, a wider range of motion is likely to occur in the actual crutching motion as compared to the more controlled motions used to characterize elbow flexion and extension properties. Additional aid to the triceps during experimentation from the shoulder musculature (or other stabilization muscles) that was previously unaccounted for in the energetics model would result in a lower empirical K_{opt}.

Finally, with more data one could potentially suggest a better curve-fitting scheme to more accurately capture the location of the optimal metabolic reduction and more carefully predict the optimal level of metabolic reduction.

5.4.3 Muscle Activation During Stair Crutching

Of all collected data, EMG is the only component that directly relates to what occurs internally at the muscular level. From the data, it is evident that that the empirical optimal location does *not* occur where the biceps and triceps activations are equal.

In addition, the sum of the muscle activations drops significantly as K_{norm} increases, Figure 5-9. This seems to support the theory that additional muscles are significant in the stair-crutching gait especially as elbow-spring stiffness becomes larger. More tests per subject (more spring stiffnesses conditions) would help more clearly define the shape of the muscle activation curves.

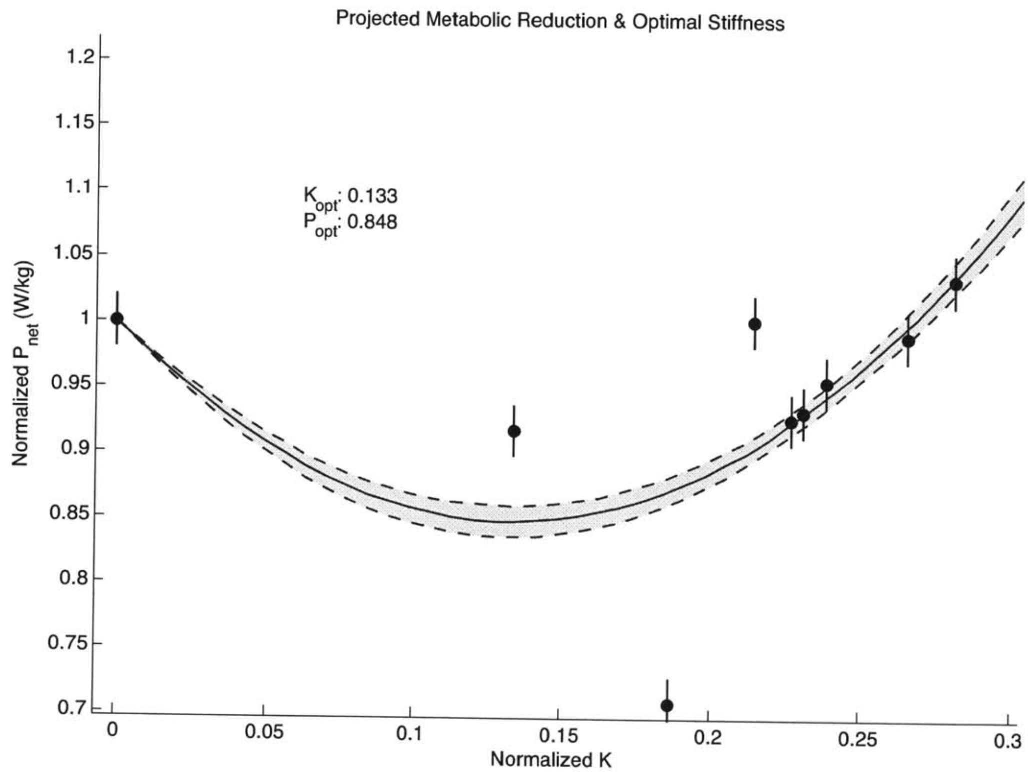


Figure 5-4: Quadratic fit to composite dataset (all subjects). The outer dotted lines represent the potential variation of the fit curve taking into account the measurement error. The center solid line is the actual best fit curve. For the purpose of this fit, only data points local to minimum are considered within a range of +/- 0.2 from the minimum datapoint. K_{opt} is the prediction of the optimal K_{norm} value for the crutching activity. P_{opt} represents the projected metabolic reduction from nominal (K = 0), in this case a 15.2% reduction.

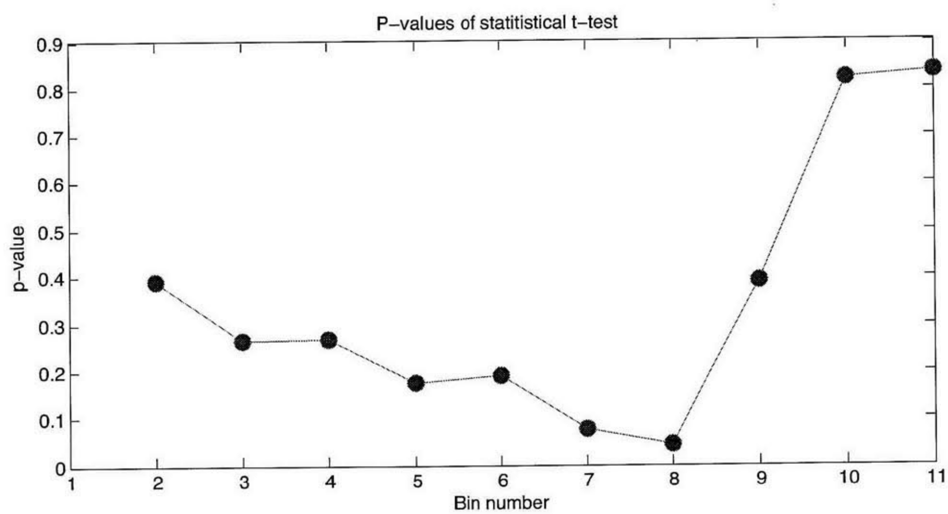


Figure 5-5: P-values from the binned statistical t-test. Each bin holds four data points. The bins correspond to sets of four data points collected sequentially along the along the K_{norm} axis. The bin with the lowest P-value is statistically significant from nominal.

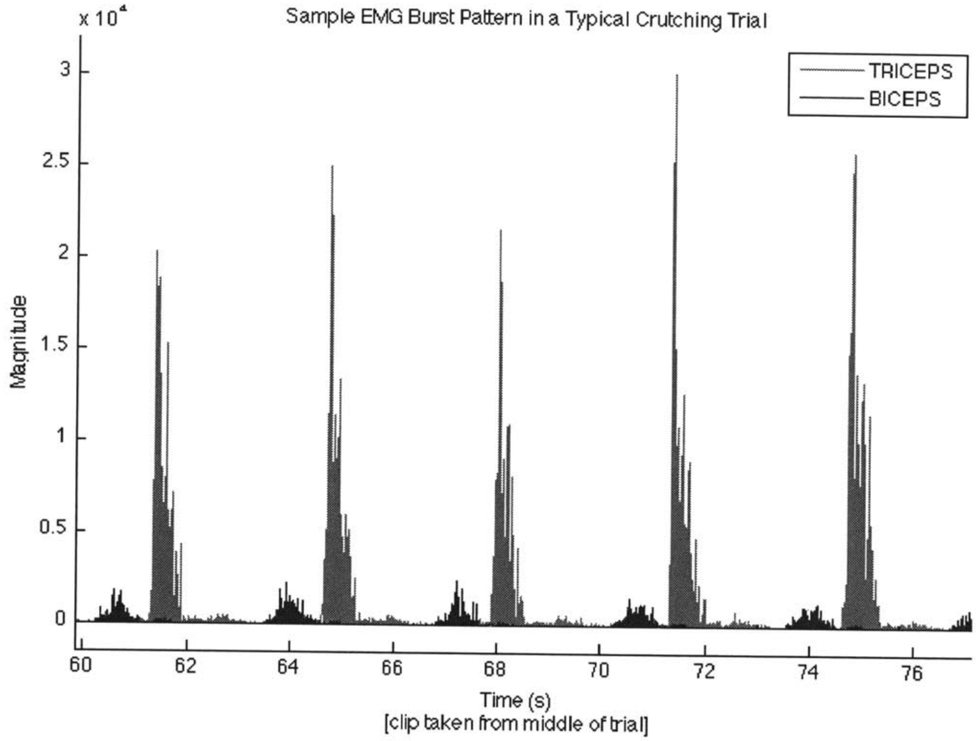
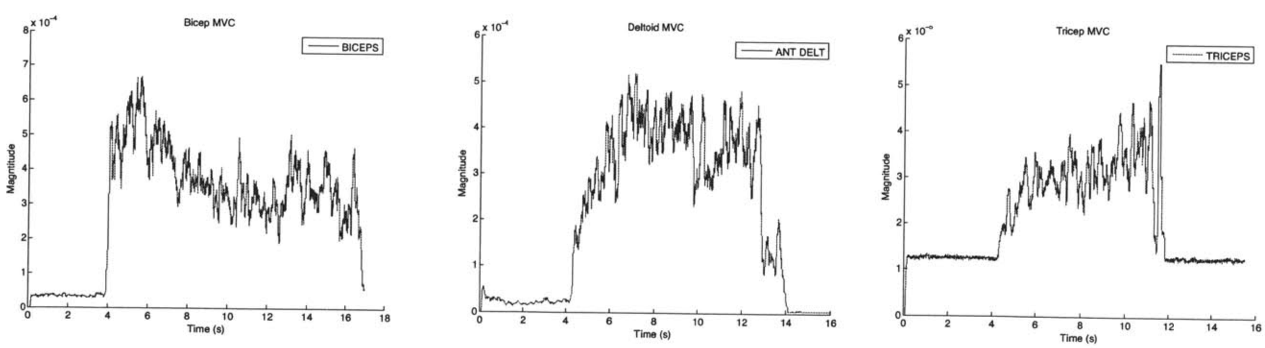


Figure 5-6: Sample un-filtered EMG burst pattern



A B C

Figure 5-7: Maximal voluntary contractions for normalizing crutching EMG. [A] Biceps MVC [B] Triceps MVC [C] Deltoid MVC

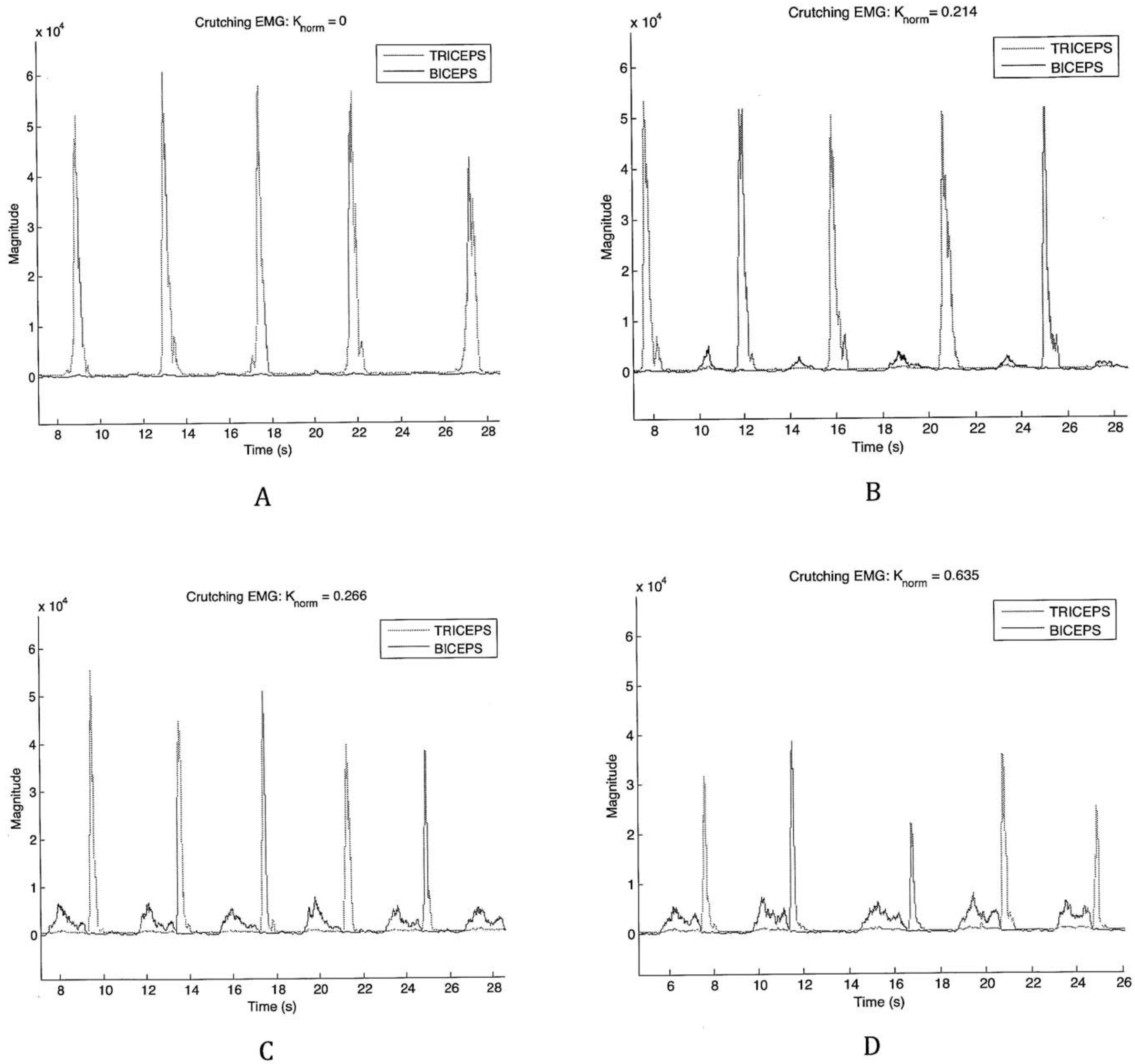


Figure 5-8: EMG readings taken during stair crutching activity with different stiffness conditions. For all figures, the x-axis reports time in seconds, the y-axis reports a magnitude in arbitrary units. [A] $K_{norm} = 0$, [B] $K_{norm} = 0.214$, [C] $K_{norm} = 0.266$, [D] $K_{norm} = 0.635$

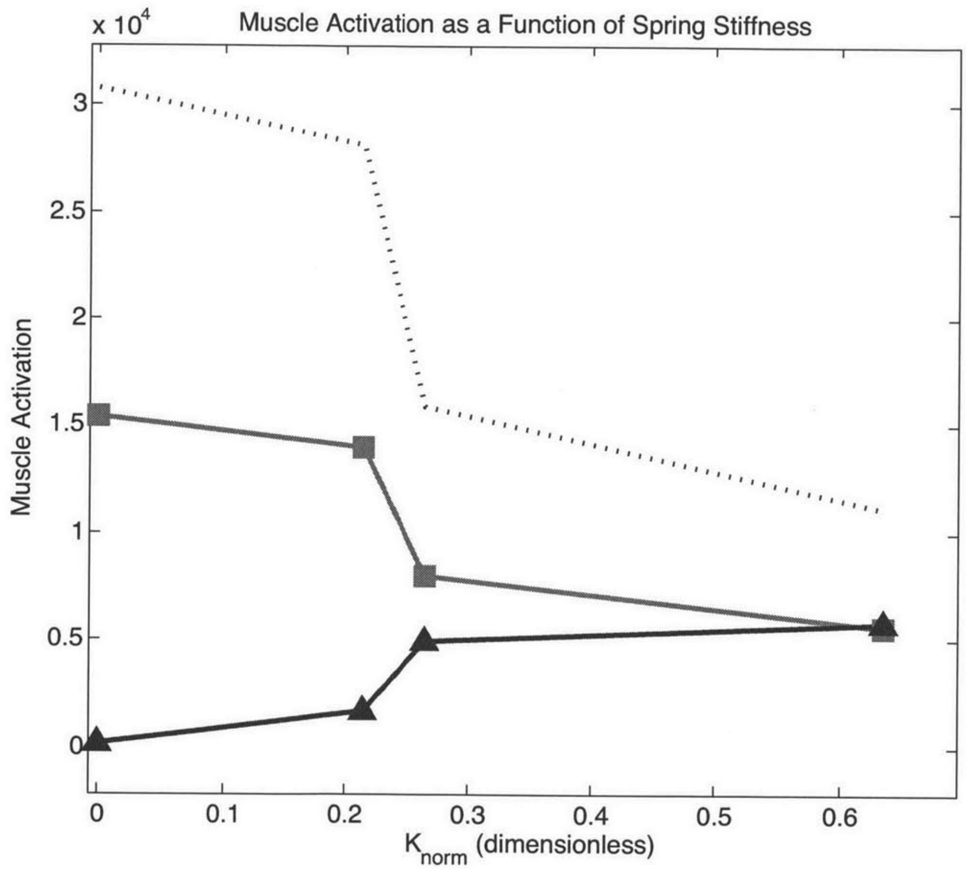


Figure 5-9: Muscle activation as it varies by spring stiffness. As the stiffness increases, the triceps activation (red squares) decreases while the biceps activation (blue triangles) increases. The dotted black line shows the sum of the biceps the triceps muscle activations.

Chapter 6

Conclusions & Future Work

6.1 Conclusion

The thesis set out to develop a tuning method by which the addition of a joint-spanning parallel spring optimizes the metabolic efficiency of movement. While framing the problem in the context of an "effective muscle", it uses empirical data and measured anatomical constants to predict and then measure an optimal stiffness for the activity of stair-crutching. Metabolic and electromyographic methods aggregate data of the whole body into one cumulative measurement while still providing insights about the inner workings of muscles under cyclic load.

The key contribution of this thesis is a the development of a crutching device which illustrates the possibility of joint tuning for metabolic efficiency. Additionally, we illustrate a rare case of augmentation where the inherent optimization of the human body is less effective alone than with the aid of a device.

The results demonstrate a trend towards significance for our hypothesis that the measured optimal metabolic efficiency can be predicted by a simple energetics model, only knowing the effective force-velocity characteristic of the joint; in this case, the elbow. While more subjects are needed to prove significance, the foundation has been laid for future data collection.

6.2 Scientific Applications

Optimization of Muscle Usage The methods of this thesis focused primarily on the collective result of metabolic reduction by the introduction of a parallel spring stiffness across a joint without looking to understand the black box that has been labeled the "effective muscle". However, one can also think about the problem in the context of a system of multiple muscles spanning the moving joints. Each of these muscle in the system has a specific fiber composition, force-velocity characteristic, and angle of origin and insertion. It is very possible that when demonstrating metabolic reduction by stiffness tuning, what is actually occurring is the collective optimization of the individual muscle force-velocity characteristics. Taking these parameters into account, one could postulate the development of a model to calculate muscle fascicle lengths during dynamic motions.

Metabolic Reduction as a Function of Speed All of the experiments in this study were performed at the same cadence. In future studies, it would be interesting to explore the concept of how metabolic reduction changes with a variation in stair-crutching speed.

6.3 Engineering Applications

The clinical results demonstrate the viability of joint stiffness tuning for the activity of stair-climbing with crutches. In principle, this method can be adapted to other joints and physical activities with a similar pattern of cyclic loading.

Exoskeleton Design Exoskeleton design is a field in which humans are augmented with mechanical apparatuses that improve their natural physical abilities. In general, while engineers have been able to greatly improve abilities like load carrying in normal subjects, they have had very little success with improving the nominal metabolic efficiency with which the body operates without such a device. Joint stiffness tuning would be helpful in this regard in activities as varies as cyclic lifting to uphill walking and running. Using the experimental methodology described here, one could impose a spring architecture and then tune accordingly. One could automate the initial tuning process by defining a series of exercises to determine the proper force-velocity properties of the joint(s) to set initial optimization bound and then actively tune to find the best stiffness while performing the requisite activity. Finally, while the optimization presented here results in optimal performance for one specific movement task, by spanning more joints or optimizing springs non-linearly, we may be able to optimize for more than one movement.

Sports & Exercise Science Just like the bicycle changed the paradigm by which sports enthusiasts could work their muscles (bicyclists can exercise over longer distances and more complex terrain made possible in part by the addition of gears which allow their leg power muscles to optimize their power point on the force velocity curve[21]), joint stiffness tuning could be useful for general exercise and weight training applications to help pinpoint a targeted muscle regime. Easily transferable activities (and their respective optimizable joint) would include: walking and running up hills (leg muscles); rowing (arm muscles), cyclic weight training.

Rehabilitation Engineering While the crutch device described here was designed for experimental use, an alternative version could be constructed for patient use. To make this device into a product, more research would be required especially focusing on the optimal design of the cuff-arm interface to make chronic cyclic loading more comfortable. Additionally, a mechanism would need to be designed to disengage the stiffness adaptation during non-optimizable activities like level ground walking. Finally, one could imagine an arm brace for wheelchair users that works on the same principle.

Appendix A

Torque-Angular Velocity Characterization Data

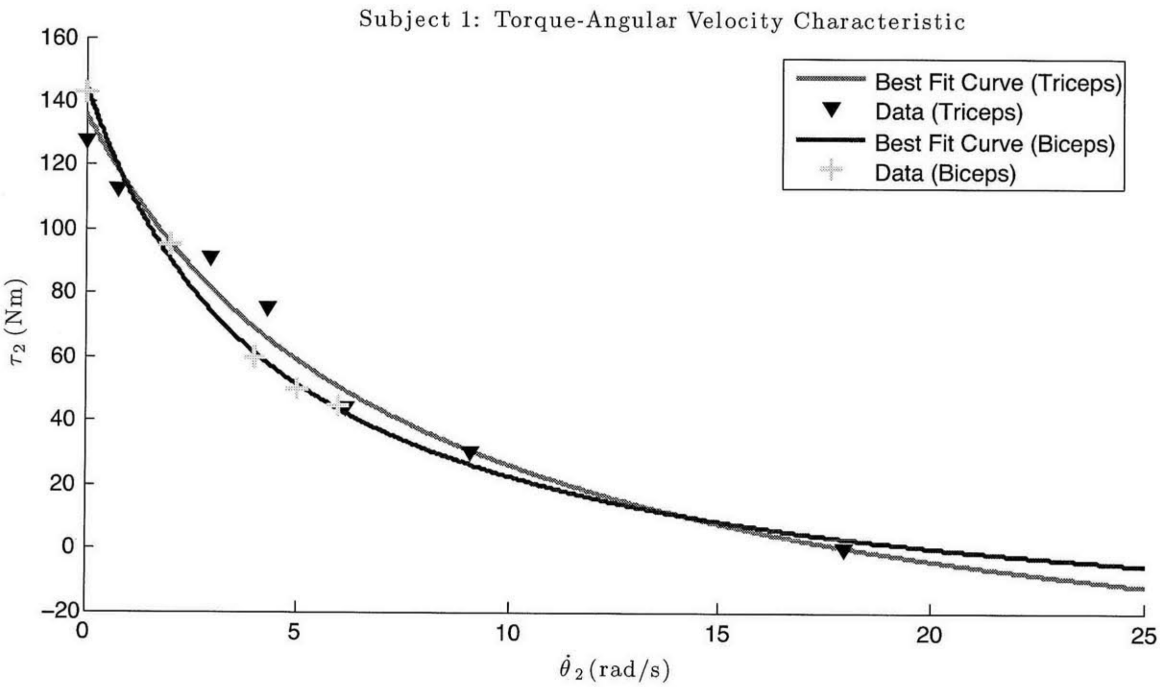


Figure A-1: A torque-angular velocity characteristic for Subject 1

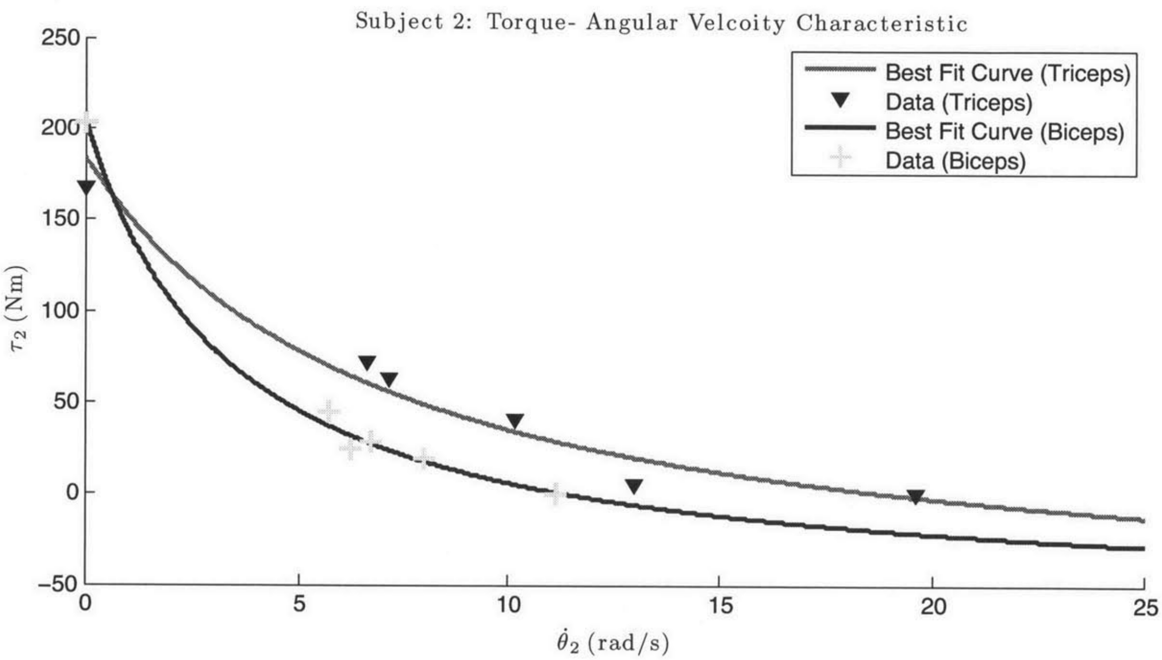


Figure A-2: A torque-angular velocity characteristic for Subject 2

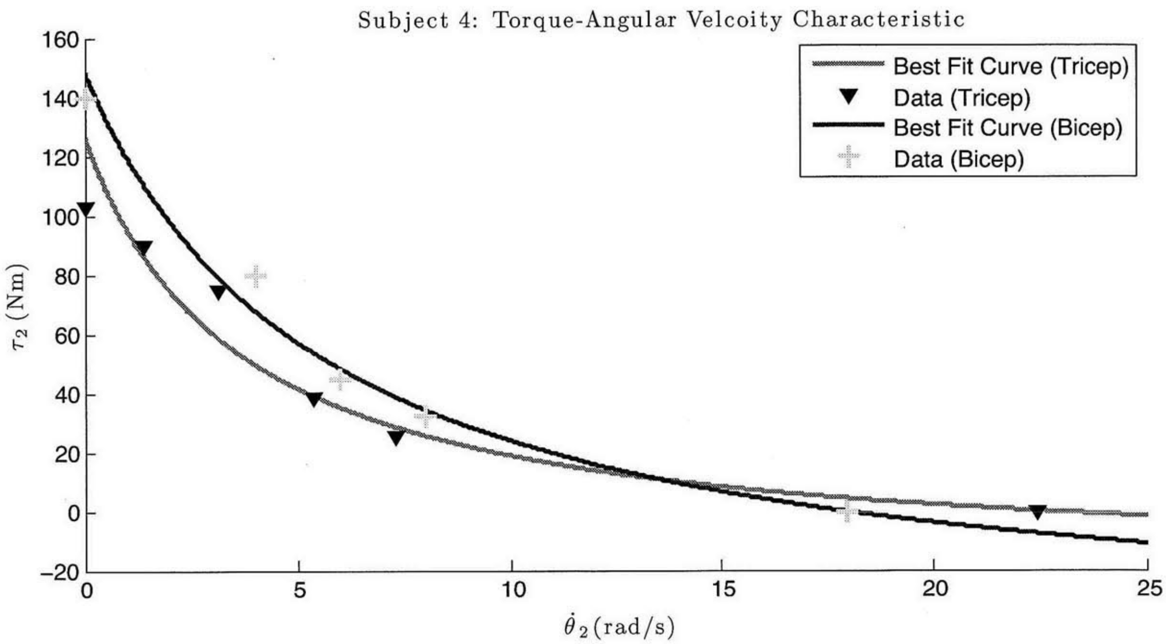


Figure A-3: A torque-angular velocity characteristic for Subject 3

Appendix B

Spring Constants

Spring Rate	OD	ID	Free Length	Solid Length	Wire Dia.	Total Coils	Material	Amt. in Series	K_{eff}
$245.2N/m$	$1.07cm$	$0.853cm$	$18.11cm$	$6.579cm$	$0.107cm$	61.8	M	1	$490.3N/m$
$875.6N/m$	$1.07cm$	$0.823cm$	$12.55cm$	$3.835cm$	$0.122cm$	31.5	MW	2	$875.6N/m$
$1086N/m$	$1.19cm$	$0.884cm$	$20.32cm$	$7.239cm$	$0.152cm$	46.5	M	2	$1085.7N/m$
$1296N/m$	$1.27cm$	$0.904cm$	$26.67cm$	$12.522cm$	$0.183cm$	68.5	M	1	$2592N/m$

Table B.1: Constants related the mechanical properties of the springs employed in this study. In 'Materials', MW refers to music wire and SPR refers to tempered spring steel.

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