IMPLEMENTATION OF SENSORS AND CONTROL IN BIOMEDICAL
REHABILITATIVE DEVICES

A Thesis Presented

by

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Abstract

Two topics were studied: impedance control was implemented in a gait rehabilitation device, and a new improved enclosure was designed for the SHIMMER wireless sensor device.

Topic 1: The field of rehabilitation robotics has been growing fast in the last several years, with devices designed to target primary gait deviations in stroke survivors and other subjects with impaired motor function. To our knowledge, no commercially available devices address secondary gait deviations in the pelvic motion. Therefore, a team of undergraduate students had designed and built the Robotic Gait Rehabilitation (RGR) Trainer II, a robotic rehabilitation device designed to interface with the pelvic area for treadmill gait retraining.

Studies have shown that the preferred method for retraining post-stroke subjects to regain motor function is by applying force fields to the targeted body components. In the field of robotics and automatic controls, force-fields are often realized by means of impedance control. Therefore, an end-point impedance controller was designed at the actuator level, and implemented in the RGR Trainer at the pelvic obliquity level. Human testing using true pelvic obliquity trajectory has shown that the system applies corrective forces to the pelvis according to the applied force-field.

Topic 2: In this design study, the plastic enclosure of the SHIMMER wireless sensor device was redesigned in order to improve its functionality. As a result, the SHIMMER’s operating time was nearly doubled, and the noise level in the acquired data was reduced.
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Chapter 1. Introduction

1.1 Motivation

Each year in the United States alone, over 750,000 people suffer strokes [1]. Stroke is the leading cause of disability [2], and about 80% of stroke victims experience weakness or trouble moving one side of their body, and require rehabilitation [3]. Gait allows individuals to perform Activities of Daily Living (ADL’s) [4] [5]. The ability to walk is also strongly correlated with the quality of life [6]. Walking and interacting with the environment may also affect brain plasticity and enhance neurorecovery [7].

The gait disorder following stroke results in hemiparesis in the affected side as well as abnormal synergy patterns. They include equines synergy, paretic synergy and reflex coactivation [8]. Stroke survivors’ comfortable walking speed (CWS) is 0.55 m/s [8] which is less than half of that seen in healthy subjects (1.52 m/s) [9]. Therefore, restoration of a normal gait pattern is a frequent goal of post-stroke rehabilitation.

Due to the factors mentioned above, a new type of robotic device for gait rehabilitation, called the Robotic Gait Rehabilitation (RGR) Trainer, was proposed by Dr. Paolo Bonato, the Director of Motion Analysis Lab at Spaulding Rehabilitation Hospital in Boston, Massachusetts. Two prototypes of the RGR Trainer had been designed and built by two teams of undergraduate students at Northeastern University in Boston, Massachusetts. The following work deals with design and implementation of impedance control in the RGR Trainer II in order to apply corrective force fields to subject’s pelvic
area and correct secondary gait deviations in post – stroke patients to augment their rehabilitation.

1.2 Background

1.2.1 Human Gait

Human gait is comprised of strides, which are the intervals between two consecutive heel strikes (Figure 1). Gait markers, (e.g. toe – off), are used to identify the phases of gait (e.g. swing phase and stance phase).

![The human gait cycle](image)

Figure 1: The human gait cycle [10]
The stance phase lasts approximately 60% of the gait cycle, while the swing phase takes up the remaining 40%. Both limbs are in contact with the ground for about 10% of the cycle, which is referred to as double limb support.

### 1.2.2 Pelvis Motion during Gait

During normal gait, the pelvis rotates in three planes: frontal, sagittal and transverse.

Figure 2: The body can be viewed in the frontal, sagittal and transverse planes [11].
Rotation of the pelvis in the frontal plane is obliquity, rotation in the sagittal plane is pelvic tilt, and rotation in the transverse plane is called pelvic rotation. These rotations happen about the supporting limb’s hip joint [12]. Pelvic drop, anterior tilt and rotation are normal events which occur during normal gait in obliquity, pelvic tilt and pelvic rotation respectively.

![Figure 3: Normal pelvis motion events in gait [12].](image)

**1.2.3 Common Gait Deviations and the Pelvis**

The most common primary gait deviation in post–stroke subjects is stiff–legged gait. This gait deviation oftentimes results in the subject employing secondary gait deviations which involve motor control of the pelvis. Stiff legged gait is associated with hip-hiking (Figure 4) or circumduction (Figure 5). Hip-hiking is an exaggerated elevation of the pelvis on the ipsilateral side (i.e. hemiparetic side) to allow toe clearance during swing, while circumduction is an exaggerated rotation of the pelvis in combination with an exaggerated hip abduction. Abnormal control of pelvic obliquity and rotation of the pelvis are the most common secondary deviations observed in post-stroke patients. A
subject will employ these secondary gait deviations in order to assist in foot clearance when either hip flexion or knee flexion are inadequate [12].

Figure 4: Hip – hike is a voluntary upward motion of the ipsilateral (affected) side of the pelvis during swing [12].

Figure 5: Circumduction is used by subjects to create additional foot clearance.
1.2.4 Gait Rehabilitation

Animal research studies have shown that goal oriented, repetitive training is the primary means of augmenting post-stroke motor relearning [13]. Human clinical trial studies that utilize goal-oriented, repetitive, active training such as constrained induced movement therapy [14], partial weight-supported ambulation [15] and robotic therapy [16] have demonstrated encouraging results. Based on the growing body of scientific evidence pointing to the effectiveness of goal-oriented motor retraining, clinicians have recently privileged a goal-oriented approach also in gait retraining, and have utilized treadmills to implement clinical protocols. Studies examining treadmill gait retraining have shown its effectiveness in improving walking velocity and other key characteristics of ambulation [15] [17] and a positive effect on mobility. Treadmill walking is used as a substitute to level ground walking since not only it appears to be an effective clinical tool, but also it offers some practical advantages over level ground gait retraining. For instance, treadmill gait retraining uses less space and is relatively simple to apply this technique in less functional patients with the use of weight support. Studies have shown that walking on a treadmill does not significantly change the gait pattern compared to level ground waking [17] and that improvements achieved during treadmill gait retraining transfer to level ground walking. In cases when patients are unable to properly ambulate, use of a treadmill for gait retraining makes it easier for physical therapists to administer motion to lower extremities manually (Figure 6).
Unfortunately, there are two major drawbacks to manual therapy: it’s difficult for the two therapists to coordinate their work properly, and it is labor-intensive, therefore making it difficult to implement in the US healthcare system.

### 1.2.5 Robotic Gait Rehabilitation

Due to the difficulties associated with manual gait retraining, robotic gait retraining systems have been developed to facilitate administration of intensive gait retraining therapy. From the point of view of training strategy and robot control, there are two types of robotic devices for rehabilitation: those which drive the body components in position mode regardless of patient efforts, and those which apply force-fields to the
body, therefore modulating the forces applied onto the body depending on patient’s efforts. The latter method, employing force-fields, has been shown to be the preferred method for retraining post-stroke subjects to regain their motor functions [18]. Therefore, only those robotic devices, which apply force-fields (with force measurement) to the lower body for the purpose of gait retraining, are presented here.

The leader in the field of robotic neurorehabilitation is a Swiss company, Hocoma AG, which manufactures the Lokomat, a robotic device for gait retraining. The system consists of the robotic gait orthosis (Lokomat), a body weight support system (Lokobasis) and a treadmill (Figure 7). This device controls the patient’s leg movements in the sagittal plane, by actuating hip and knee joints. The force fields are realized by use of impedance control. The device also features a passive foot lifter, which helps with ankle dorsiflexion in the swing phase. The subject’s pelvis is fixed in the horizontal plane, but slight rotations of the pelvis are possible due to cushions and straps used to hold the body [19]. With the body weight support included, the Lokomat system features five degrees of freedom (all actuated). This system is commercially available.
Another robotic device, which is designed to apply force fields in gait retraining, is the LOPES, from University of Twente in The Netherlands. This device is similar to the Lokomat, but in addition to controlling the hip and knee joints in the sagittal plane, the LOPES features additional degrees of freedom to allow pelvis translations in the horizontal plane, as well as hip abductions (Figure 8). The device is not available commercially.
Figure 8: LOPES’s nine degrees of freedom (eight are actuated) [21].

Figure 9: LOPES [21].
Another device, which employs force feedback in application of motion trajectories to the lower body, is the HapticWalker from the Fraunhofer Institute for Production Systems and Design Technology in Berlin, Germany.

This device is comprised of two 3 degree-of-freedom modules, which use 6 DOF force/torque sensors in its foot plates. Up to six DOF’s per foot are available. The unique design of the HapticWalker allows for simulation of a wide number of trajectories like stair climbing, but it uses position control. The latest published article about this device [23] mentions force-field type control algorithms being under development.

The Lokomat, the LOPES and the HapticWalker systems described above have the capability to correct primary gait deviations, such as knee hyperextension during stance
and stiff legged gait defined as limited knee flexion during swing, but the secondary gait deviations in the pelvis are not targeted. Thus, the RGR Trainer was proposed by Dr. Paolo Bonato, to facilitate robotic gait retraining using force-fields applied to the secondary gait deviations in pelvic motion.

1.3 Overview of Previous Work

1.3.1 Introduction

The second prototype of the Robotic Gait Rehabilitation Trainer (RGR Trainer II) came into existence in December of 2006 as a result of the work of two undergraduate Capstone Project teams at Northeastern University’s College of Engineering. This prototype was designed with control of one degree of freedom in mind – obliquity, while the remaining five degrees of freedom are non-actuated. They are: pelvic rotation, pelvic tilt, side-to-side translation, forward-back translation and up-down translation. Hardware for an un-weighing system was also designed.
1.3.2 Machine Design

A CAD model of this prototype is pictured in Figure 11.

Figure 11: CAD model of the RGR Trainer II, as designed by the second Capstone team, with the actuated degree of freedom (obliquity) shown. The treadmill is not pictured.

In RGR Trainer II, the motive power comes from two linear servo tube actuators from Copley Controls Inc. (Canton, MA, USA). The generated forces are transferred to the pelvic area via the Newport 4 pelvic brace from Orthomerica Products Inc. (Newport Beach, CA, USA). The design was based around a Biodex un-weighing device’s frame (Shirley, NY, USA). As is shown in Figure 11, the two linear actuators were attached to this frame with two intersecting mechanical slides on both sides of the body in order to allow motion in the non-actuated degrees of freedom.
1.3.3 Sensor Integration

For the purpose of obtaining force feedback, a compression-tension model 31 load cell from Honeywell (Morristown, NJ) was placed in an assembly comprised of two aluminum bars connected at one end by a hinge. The purpose of the assembly was to isolate axial loads from any side loads and bending moments. The assembly was placed at the interface between the actuator’s thrust rod and the pelvic brace, connected via a spherical joint, as shown in Figure 12.

![Figure 12: Force field mechanism details.](image)

Position feedback (of the actuator thrust-rod) was acquired directly from the Xenus servo-amplifier, which controlled one of the linear actuators (only one servo-amplifier was available).
1.3.4 Control Hardware and Software

The linear servo-tube actuators used require a sophisticated digital Xenus servo – amplifier for control. This unit can be configured to operate in position, velocity or force mode. Here, the force mode was used. Xenus features a built-in unity feedback current loop. An internal sensor monitors the current consumed by the actuator, and a PI (proportional and derivative) controller adjusts the voltage sent to the actuator in order to coax the requested current draw (which is equivalent to force command). The block diagram of the amplifier’s internal control loop is pictured below.

Figure 13: Outline of the inner current loop contained in the Copley Controls Inc. Xenus servo amplifier.

The higher-level controllers were realized using LabVIEW software (National Instruments Inc. Austin, TX), on a Pentium 4 desktop PC and Windows OS. The force feedback was acquired with an A/D data acquisition card, while the communication with the Xenus amplifier (position feedback and force commands) was carried over a CANopen communication protocol, which used a Copley Controls Inc. PCI CAN card. This hardware allowed for the control loop to operate at about 200Hz (5ms period).
1.3.5 Testing

The device was tested in three modes: passive, back-driven and active. In the passive mode, subject’s motion trajectories were captured. Then, the trajectories were captured in the back-driven mode. The captured trajectories were played back in the active mode by employing the controller shown in Figure 14. Also, foot switches were used to test the feasibility of man-machine synchronization.

![Control architecture used by the Capstone Team to realize force fields.](image)

Figure 14: Control architecture used by the Capstone Team to realize force fields.

1.3.6 Results

The students’ efforts resulted in a second prototype of the RGR Trainer, which uses highly back-drivable actuators. The mechanism was designed with high rigidity and low friction in mind. This mechanism together with the actuators made it possible to apply force fields to the pelvic area to influence obliquity. Unfortunately, due to time constraints and other limitations, little was accomplished in terms of designing proper, stable control. Force feedback proved to be unreliable, and the impedance control
algorithm from Figure 14 behaved in a stable manner only at very low gains. Therefore, it was only possible to exert low corrective forces onto the subject’s pelvic area. In addition, while the weight support system allowed for movement in the horizontal plane, the support was passive, with no active force or position adjustment.
Chapter 2. Improvements to RGR Trainer II

2.1 Introduction

Due to difficulties encountered by the capstone team, high-strength force fields had yet to be applied to a subject’s pelvic area while walking, in order to confirm the feasibility of such approach to target secondary gait deviations, and obliquity in specific. Therefore, as the project was revived in August 2008, the main goal set forth was to conduct human testing of the RGR Trainer II and answer this and other questions. Based on the results of preliminary testing conducted by the capstone team, it became clear that the system required improvements in most areas in order to be used to verify the assumptions set forth when the project was conceived. These areas were: mechanism design, control hardware and software improvements, controller design, implementation of safety features and others.

2.2 Mechanical Design

2.2.1 Actuator Assembly Redesign

Each servo-tube actuator was positioned vertically in the device, and the two slide assemblies per each actuator allowed for movement in the horizontal plane (Figure 15). The servo tube linear actuator was designed to apply forces in the axial direction (here the vertical direction), and in fact the thrust rod was not meant to take significant side-loads. There exists an air-gap between the actuator body and the thrust rod, which is unguided. Therefore, side-loads caused the thrust rod to rub against the inner polymer
lining of the actuator body, resulting in intermittent and unpredictable static and Coulomb friction. In addition, the load-cell sensor was not located axially with respect to the thrust rod, as shown in Figure 16. Both of the mentioned design features caused unwanted friction and unpredictable resistance, which were reported by subjects using the device as “mechanism motion not smooth”.

Figure 15: Original design of actuator – pelvic brace interface.

Figure 16: Original load cell assembly mounted between thrust rod and pelvic brace. The spherical joint is offset from centerline of thrust rod, causing unpredictable friction variations.

In order to alleviate the mentioned issues with the original design, a new actuator assembly was designed, as shown in Figure 17. It consists of a stock aluminum channel,
which serves to align two pillow blocks below the actuator, with their linear bearings located concentrically to the actuator. Since the thrust rod contains a stack of high-power magnets, the shaft could not be guided directly by steel ball bearings. Polymer bearings had been found to cause too much friction under side-loads. Therefore, a ½ inch diameter case-hardened steel shaft was used to extend the thrust rod. This steel shaft was supported by the two linear ball bearings mentioned earlier. Two adjustable limit stops were realized by placing two aluminum collars on the ½ inch shaft within the assembly. This protected the bearings, and doubled as a safety feature. The load cell was placed directly at the end of the ½ inch steel shaft, with a spherical joint at the opposite end interfacing with the pelvic brace. Finally, both for safety reasons and to increase rigidity of this assembly, a clear Plexiglas guard was used to keep pinch-points out of reach. This assembly was mounted in the device above the pelvic brace, with the end of arm reaching downward to interface with the brace (Figure 18). This modification was put in place based on feedback from subjects, who noted that the original orientation of the actuator restricted range of motion of lower limbs in abduction.
Figure 17: Redesigned actuator assembly.

Figure 18: CAD model of the redesigned actuator assembly mounted upside-down to allow for leg abduction.
2.3 Mechanism Friction Measurement

2.3.1 Introduction

As it was mentioned earlier, RGR Trainer II was designed to allow for application of force fields in just one degree of freedom – obliquity. Therefore, the device was designed to allow close to unrestricted motion in the remaining non-actuated degrees of freedom. Despite the efforts taken, mechanical components exhibit certain inherent static and Coulomb friction, and these reduce backdrivability. Design features like wipers, which are meant to keep the precision parts from contamination with fine particles, can add significant friction to linear guides. Since relatively large components were used to minimize play and deformation, the resulting frictional forces affecting the non-actuated degrees of freedom can be quite significant. For the purpose of quantifying these resistive forces, and to generate a point of reference for future redesigns, the friction in the slide assemblies was measured experimentally.

2.3.2 Friction Measurement Procedure

The actuation mechanism was configured to reflect the de-powered device conditions. In an unpowered state, only the actuator housing is supported by the linear slides, while the thrust rod’s weight is supported by the subject.
Figure 19: The actuator supported by mechanical slides moves in the directions indicated (forward-back and side-to-side).

Model TMD01 electromechanical linear motion actuator from Duff-Norton (Charlotte, NC, USA) was used to set the linear guide and the telescoping rail in motion. The static friction force of Thomson linear guide 15C 460L (Danaher Motion, Radford, VA, USA) and the Rollco ASN43-770 telescopic rail (Jhong He City, Taiwan) was measured with a Futek QLA216 load cell (Irvine, CA, USA). In order to capture the maximum friction force, the supply of voltage to the Duff-Norton actuator was slowly increased, while a National Instruments data acquisition system and LabVIEW software (Austin, TX, USA) were used to record the force readings. Four such tests were performed. Load cell data was post-processed by applying a 30 point moving average. The test producing the lowest magnitude force peak was chosen, having assumed that this reading was least influenced by inertia.
Figure 20: Static friction force in one Thompson linear guide was quantified at 0.78 N. The same procedure was used to find static frictional force in the Rollco telescoping slides.

Figure 21: Static friction force in a pair of Rollco telescoping slides was measured to be about 18.2 N.

Next, dynamic friction forces were measured in the two directions of motion. Coulomb friction was measured at four different velocities by operating the Duff-Norton linear
actuator at four different voltages: 6, 8, 10 and 12V. A linear potentiometer was used to measure the resulting velocities, and the same load cell as before was used to measure the forces necessary to produce the motion.

Figure 22: Force and displacement data was used to find Coulomb friction. The results of Coulomb friction force measurements are summarized in Figure 23.

Figure 23: Coulomb friction in Thomson guide (fore-aft) and Rollco rails (sideways).
2.3.3 Friction Measurement Results

Measurement of friction forces in the actuator support mechanism allowed for quantification of resistances, which a subject experiences when the RGR Trainer II is unpowered. These resistances depend on the subject’s motion, and are summarized below.

Table 1: Resistance to movement in the unpowered RGR Trainer II. For pelvic rotation, a 56cm attachment span was assumed.

<table>
<thead>
<tr>
<th>Motion Type</th>
<th>Static</th>
<th>Dynamic</th>
</tr>
</thead>
<tbody>
<tr>
<td>Forward-Back</td>
<td>1.56 N</td>
<td>1.2 N (at 0.035 m/s)</td>
</tr>
<tr>
<td>Side-to-Side</td>
<td>36.4 N</td>
<td>25.2 N (at 0.035 m/s)</td>
</tr>
<tr>
<td>Pelvic Rotation</td>
<td>0.88 N-m</td>
<td>0.67 N-m (at 0.125 rad/s)</td>
</tr>
</tbody>
</table>

As we can see in Table 1, the low frictional forces in the Thomson linear guides result in low resistance imposed on the subject when moving forward-back (only 1.56 N). At the instant the subject is positioned perfectly straight ahead, resistance to pelvic rotation is also very low (0.88 N-m static), since only the Thomson linear guides are engaged. The values for pelvic rotation listed in the table are the minimum resistive torques. Once the subject rotates his/her pelvis to a position different than “straight on”, the Rollco rails are engage into motion. These rails have high inherent friction, resulting in about 36.4N of resistive force to side-to-side motion. This high friction, depending on the pelvis’ orientation, causes high resistive torques to pelvic rotation.

The data obtained here can serve as a benchmark for assessing backdrivability of a new design.
Chapter 3. Impedance Control

3.1 Introduction

In order to correct secondary gait deviations in subjects, force fields will be applied to the subject’s pelvic area. The force fields will be realized in the physical sense by use of impedance-controlled linear actuators, which will transfer forces and/or moments to the subject’s pelvis via pelvic brace.

3.2 PD Position Control

A simple way to apply force fields to an object is by use of P (proportional) only, or PD (proportional and derivative) - controlled actuator/robot. With a highly backdrivable actuator, the proportional gain has the effect of a spring’s stiffness constant K, and the derivative gain acts like the damping constant B of a physical damper [24]. Stiction, Coulomb friction and inertia greatly affect the behavior of such a system, reducing its ability to display desired endpoint impedance [25].

3.3 Impedance Control Overview

Here, when speaking of impedance control, we refer to the control of the end-point impedance of a robot or an actuator. Impedance control architecture consists of an inner unity feedback force loop, and an outer unity feedback position loop. The main task of the force loop is to increase backdrivability of the actuator. In that sense, force feedback moves any actuator closer to an ideal source of force. The outer position loop sets the relationship between the position of the end-effector, and the force it exerts. This is
usually accomplished with a PD controller, where the proportional term represents virtual spring stiffness, and the derivative term acts like a virtual damper. A simple schematic of an impedance controller is shown in below.

![Figure 24: Basic outline of impedance control architecture.](image)

In order to compute the transfer function of the above control scheme, we assign certain properties to the environment: $B_e$ and $K_e$. This can be represented with the block diagram below:
This block diagram describes the interaction between the robot/actuator and the environment. The inner-most loop says that the actuator end point acceleration is

$$\ddot{X} = \frac{F_{\text{act}} - F_{\text{ext}}}{m_{\text{act}}}.$$  

The acceleration of the end effector depends on the force applied by the actuator and the dynamics of the environment, which produces an opposing force. The transfer function of the diagram in fig. 2 is:

$$\frac{F_{\text{act}}}{X_{\text{ref}}} = \frac{G}{G+1} (B_c s + K_c) \left( m_{\text{act}} s^2 + B_e s + K_e \right) \frac{m_{\text{act}} s^2 + s \left( B_e + \frac{G}{G+1} B_c \right) + \left( K_e + \frac{G}{G+1} K_c \right)}{G+1} $$  \hspace{1cm} (3.1)

This transfer function above demonstrates the complexity of the dynamics involved in impedance control.
3.4 Effect of Force Feedback on Actuator Mass

The analysis presented in here is based on Hogan’s work in [26].

Neglecting friction, the actuator’s thrust rod can be represented as a mass \( m \) undergoing displacement \( x \) due to forces \( F_{\text{act}} \) applied by the actuator’s electromagnetic field, and \( F_{\text{ext}} \), or external force, applied by the environment.

\[
m_{\text{act}} \ddot{x} = F_{\text{act}} - F_{\text{ext}} \tag{3.2}
\]

\[F_{\text{act}} = G(F_{\text{ref}} - F_{\text{ext}}) \tag{3.3}\]

These two equations combined give us the following equation:

\[
m_{\text{act}} \ddot{x} = G(F_{\text{ref}} - F_{\text{ext}}) - F_{\text{ext}} \tag{3.4}
\]

And the transfer function is:
This can be represented by the block diagram below:

![Block Diagram](image)

Figure 27: Actuator shaft and force control law.

The immovable mass (body) with stiffness and damping, with $F_{\text{ext}}$ being the interaction force between the body and the actuator, can be represented by the first order equation:

$$F_{\text{ext}} = B_e \dot{x} + K_e x$$  \hspace{1cm} (3.6)

And its Laplace is:

$$X = \frac{F_{\text{ext}}}{B_e s + K_e}$$  \hspace{1cm} (3.7)

Now we combine the actuator TF with the body’s TF to describe the actuator – body interaction:

$$\frac{F_{\text{ext}}}{F_{\text{ref}}} = \frac{(B_e s + K_e) G}{(G+1)(G+1) s^2 + B_e s + K_e}$$  \hspace{1cm} (3.8)

Where $\frac{m_{\text{act}}}{(G+1)}$ is the apparent inertia as experienced by the environment. Therefore, the effect of force feedback is the reduction of the apparent actuator inertia by a factor of $G+1$. 

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3.5 Derivation of End - Point Impedance Controller

In [26], Hogan presents an impedance controller for stable contact execution between a robot and the environment. The following impedance controller derivation is an adaptation of Hogan’s work for controlling actuator’s end point impedance in the RGR Trainer.

The simplified actuator dynamics are shown in the figure below.

![Dynamics of the actuator.](image)

Figure 28: Dynamics of the actuator.

The equation describing the dynamics is:

\[ m_{act} \ddot{x} = F_{act} - F_{ext} \quad (3.9) \]

The force generated by the actuator onto the thrust rod:

\[ F_{act} = m_{act} \ddot{x} + F_{ext} \quad (3.10) \]

The desired end-point impedance of the actuator thrust rod can be represented by the following equation:
\[ F_{\text{ext}} = M_c (\ddot{x}) + B_c (\dot{x}_o - \dot{x}) + K_c (x_o - x) \]  \hspace{1cm} (3.11)

And the desired acceleration of the actuator thrust rod is:

\[ \ddot{x} = \frac{1}{M_c} [K_c (x_0 - x) + B_c (\dot{x}_o - \dot{x}) - F_{\text{ext}}] \]  \hspace{1cm} (3.12)

Now substitute the desired acceleration into the actuator force equation:

\[ F_{\text{act}} = \frac{m_{\text{act}}}{M_c} [K_c (x_0 - x) + B_c (\dot{x}_o - \dot{x}) - F_{\text{ext}}] + F_{\text{ext}} \]  \hspace{1cm} (3.13)

And

\[ F_{\text{act}} = \frac{m_{\text{act}}}{M_c} [K_c (x_0 - x) + B_c (\dot{x}_o - \dot{x})] + F_{\text{ext}} [1 - \frac{m_{\text{act}}}{M_c}] \]  \hspace{1cm} (3.14)

The equation above describes the impedance controller. \( F_{\text{act}} \) is the force commanded to the servo-amplifier. Ideally, the inertia of the thrust rod mass - \( m_{\text{act}} \), as seen by the patient, should be minimized. In practice, this apparent inertia can only be reduced to a certain degree by use of force feedback. Therefore, here we equate the desired mass \( M_c \) to the lowest possible apparent inertia of the thrust rod: \( M_c = \frac{m_{\text{act}}}{(G+1)} \) and the force controller gain \( G \) is picked to be highest possible, while still providing appropriate stability margin. After the substitution, the equation describing force commanded to the actuator \( F_{\text{act}} \) is:
\[ F_{\text{act}} = (G + 1)[K_c (x_0 - x) + B_c (\dot{x}_0 - \dot{x})] - (G)F_{\text{ext}} \quad (3.15) \]

The above equation lists the constituents of the force command \( F_{\text{act}} \), which is fed into the servo amplifier, to be executed by the actuator. This can be represented by the following diagram:

![Diagram of physical implementation of equation (3.3).](image)

Figure 29: Physical implementation of equation (3.3).

The output of the PD controller, which acts on the position error, can be called the virtual force, \( F_{\text{virt}} \). It is the output of the virtual spring’s and virtual dampers stiffness, \( K_c \) and damping \( B_c \) respectively.

It is a known fact that force controller gains are often limited to single digits. At such low gain values, the steady state error can be very significant. For example, using the control law of equation (3.3) and a proportional gain \( P=1 \), the resulting force output \( F_{\text{ext}} \) is only 50% of the input \( F_{\text{ref}} \). The impedance controller from Figure 29 takes this effect into account, magnifying the PD controller’s output by \( (G+1) \) to cancel the steady state error resulting from the control law and low gain value. Due to force feedback’s dependence on the environment, tuning is often performed manually [25], as we will see in the next chapter.
Chapter 4. Physical Implementation of Impedance Control

In the previous chapter an end point impedance controller for the servo tube linear actuator was derived. In this chapter the implementation of such a controller in hardware and software is described.

4.1 Control Hardware and Software

The control hardware consists of a 6259M PCI data acquisition card (DAQ) from National Instruments (Austin, TX, USA), which runs in LabVIEW Real-time operating system (RTOS) on a dedicated PC (target). A host computer serves as the user interface.

Windows operating systems were designed to run various operations simultaneously and provide processing power to any application which requests it. The result is a non-deterministic performance, since the high-priority feedback control loop operation can be interrupted by peripherals, such as the mouse and keyboard. This is in contrast to real-time operating systems, which are designed to complete operations within a specified amount of time, and such performance is called deterministic [27].
4.2 Actuation Hardware

The servo-tube actuator (model STA2504) from Copley Controls Inc. (Canton, MA, USA) is direct-drive electromagnetic linear motor, with windings in the actuator housing, and permanent rare-earth magnets in the movable thrust-rod (Figure 31). The servo tube is a very good source of force and lends itself very well to impedance control [25]. Its total mass is 3kg, while the inertia of the moving forcer (the thrust-rod) is 1.67 Kg, with a 25cm stroke length. This actuator can output 51N continuously, and up to 312N peak force (for 1 second). In some applications like exoskeleton actuation, this linear motor’s force density (ratio of force output to inertia) may be too low, but in the RGR Trainer, where the actuator body is supported and not subject to large movements, this choice seems appropriate.
The servo tube linear actuator requires the Xenus servo amplifier for operation. This digital amplifier can be programmed to operate in three different modes: position, velocity or force.
In the force-mode, the Xenus servo-amplifier does not use a direct measure of force, but it does however monitor the current consumed by the actuator, and a proportional-integral (PI) controller adjusts the voltage sent to the actuator in order to coax the requested current draw (which is equivalent to force command). The block diagram of the amplifier’s internal control loop is pictured below. An automatic tuning procedure performed by the Xenus servo amplifier set the current loop gains to $C_p = 454$ and $C_i = 88$.

![Block diagram of the amplifier’s internal control loop](image)

Figure 33: Outline of the inner current loop contained in the Xenus servo amplifier.

### 4.3 Force Feedback

It was shown in the previous chapter how the equation defining end point impedance of an actuator is derived. The degree, to which the actuator system can actually display the specified endpoint impedances, depends largely on the extent of backdrivability of the actuator. The higher backdrivability, the closer the commanded impedance matches the actual value. Therefore, proper implementation of force feedback is crucial to implementation of impedance control.
4.3.1 Load Cell Signal Conditioning

In order for a load cell signal to be useable, the noise inherent to force signal needed to be attenuated. The noise was first quantified by commanding the actuator with the included CME 2 software to apply cyclical sinusoidal loading (amplitude ~31.2N). The signal from the Honeywell-Sensotec model 31 compression – tension load cell was amplified by the Honeywell-Sensotec UV-10 in-line amplifier. The signal was acquired with National Instruments 6259M data acquisition card at 2000Hz and LabVIEW Real-Time. A sample of the unconditioned data is shown below.

![Unconditioned Load Cell Sample Data](image)

Figure 34: The unconditioned load cell signal contains significant noise.

Next, an analog anti – aliasing low pass RC (1 pole Butterworth) filter was placed just aft of the amplifier, with a cutoff frequency set to 480Hz. Another data sample was collected, a portion of which is presented below.
Figure 35: Once analog – filtered, the noise level in the signal is greatly reduced. Next, with help from Dr. Paolo Bonato from Spaulding Rehabilitation Hospital in Boston, Massachusetts, power spectrum density graphs were generated, in order to confirm noise attenuation and help design a digital filter.
As can be seen from the graphs above, many of the signal frequencies, which existed in the unfiltered signal in the 0 to 480Hz range were removed by the low pass filter. This suggests that the removed signal components were aliases of higher frequency signal
components. Next, a digital filter was designed in LabVIEW. A target cutoff frequency of 8Hz was chosen, based on the frequency range of interest between 0 and 8Hz.

![Graph of Analog and Digital - Filtered Load Cell Data](image)

Figure 38: Combination of analog – filtering to remove high frequency aliases, and digital filtering produces a clean signal.

### 4.4 Position and Velocity Feedback

In order to conduct measurement of environment dynamics presented above, and to control end-point actuator impedance in general, it was necessary to have thrust rod’s position and velocity signals. Copley’s servo tube actuators are equipped with hall sensors, which are used by the Xenus servo amplifier to generate an emulated differential quadrature encoder signal (position). Since our DAQ card can only acquire single-ended encoder signals, the emulated differential signal from Xenus was converted to single ended using a US Digital Inc. incremental encoder adapter. The encoder signal is
acquired by the National Instrument Inc. 6259M DAQ card, where a hardware counter operates at 80MHz, counting both rising and falling edges of the incremental encoder signal (X4 encoding). The net number of counted edges is polled by the controller at the same frequency as the controller’s operating rate, (here 10 kHz) and converted to position with knowledge of encoder’s resolution. With the amplifier’s emulated incremental encoder signal resolution of 12.5 microns, the result is very high quality digital position feedback.

For the purpose of obtaining velocity feedback, the acquired position signal is differentiated in LabVIEW. The graph below shows sample result of the operation. The counter was polled at 10 kHz, and the data was down-sampled to 2 kHz for display purposes.

Figure 39: Position signal and its derivative, acquired at 2 kHz. The shape of the derivative curve is a result of discretization.
In order to be able to use such a signal, it has to be averaged first, to produce a smooth curve. A moving average was implemented for that purpose, using 75 points. The resulting curve is shown below.

![Velocity - Moving Average](image)  

**Figure 40:** A 75 point moving average of the velocity signal resulted in a relatively smooth curve, incurring a 3.75 millisecond delay (based on 10 kHz acquisition rate).

### 4.5 End – Point Impedance Controller – System Architecture

The end point impedance controller was first implemented at the actuator level (linear motion). The system diagram of equation (3.3) specifying actuator force $F_{act}$ is shown in Figure 41.
Figure 41: Impedance controller at actuator level - system diagram.

The actuator position feedback and the reference trajectory generate an error in position. The position error and its derivative are multiplied by the virtual stiffness and virtual damping constants, $K_c$ and $B_c$ respectively, to produce a “virtual force” $F_{\text{virt}}$. As dictated by equation (3.3), the $F_{\text{virt}}$ term is multiplied by the $(G+1)$ term to produce the reference force signal, $F_{\text{ref}}$. This term is compared to the product of force feedback and the force loop gain $G$. The difference between the two terms, $(F_{\text{act}})$ is sent to the Xenus servo amplifier as the force command. A diagram of physical hardware and connections between them is shown in Figure 42 below.
4.6 Safety

4.6.1 Problem Statement

The Xenus servo amplifier employs a Schmitt trigger in its enable function to recognize the “enable” signal (greater than 3.65V) and the “disable” signal (less than 1.35 V). In general it is desirable to ensure that the drive be disabled when the control software or the computer itself fail. A simple solution of using a DAQ analog output channel to send an “enable” signal while the controller is in operation, and a disable signal when it stops is not perfectly reliable. Unfortunately, when LabVIEW software fails, the National Instruments 6259M card continues to output the last value for as long as there is power.
supplied to the computer. This creates a potentially dangerous situation, especially with a highly backdrivable system as ours working under impedance control. For example after a software failure, a high-force command from the time of failure will remain. Physically unplugging the servo amplifier’s mains electrical supply would cause the actuator to cease exerting force thus becoming easily backdriven to a new position. Then, returning mains power to the amplifier could cause a sudden, unexpected acceleration of the thrust rod, possibly causing injury and damage to the actuator assembly.

4.6.2 Safety Solution

To address this problem, one solution is to use an external microchip with a “watchdog” feature, which detects software failure, as was done in [22]. Our solution was to use a simple analog circuit to provide the enable signal to the Xenus servo amplifier only when LabVIEW is active. A dedicated DAQ output was configured to supply a sinusoidal voltage signal of 100Hz frequency and ranging between 0V and 10V. This signal was routed through an analog RC high-pass filter, with the cutoff frequency on the order of several Hz to avoid excessive signal attenuation. Then, the signal was rectified with a Gratz bridge rectifier and smoothed with help of a capacitor placed in parallel, as is shown in Figure 43. The result was a slightly varying voltage output which successfully enabled the Xenus servo amplifier when the input was of proper frequency and magnitude. At the same time, the circuit’s output changed to 0V whenever the input was 0V or un-varying (as in the case of software error). Throughout bench testing, this scheme has been proven to work very well.
Figure 43: Analog amplifier-enable safety circuit.
Chapter 5. Force Fields Applied to the Body using Impedance Control

5.1 Introduction

As a stepping stone to using our end point impedance-controlled servo tube actuator to generate force fields around the pelvis at the obliquity level, the system was bench tested and its performance was characterized.

5.2 Force-Loop Tuning

5.2.1 Introduction

In order to apply force fields using impedance control to the pelvic area required that the force loop be tuned for that specific environment. Therefore a set of tests were designed and carried out, and a range of acceptable force loop gains was found. The impedance control architecture derived in chapter 3 uses an inner force loop. Since the performance of force feedback depends largely on the environment, gains are often found by manually tuning the force controller [25]. Therefore the closed loop force controller was tuned for optimal performance. The end point impedance controller derivation presented in Chapter 3 assumed a proportional force loop gain only. Therefore, only proportional controllers were tested.
5.2.2 Test Protocol

The servo tube actuator assembly was positioned vertically and attached to a test frame. The subject wore the Newport 4 pelvic brace with a thigh segment on the left leg. The actuator assembly’s guide shaft was connected to the pelvic brace through the compression–tension load cell and a spherical rod end. A ¼-28 threaded rod connected the brace to the spherical joint. The subject remained still, while a step input of 50N was introduced to the amplifier using LabVIEW, and the resulting force between the actuator shaft and the pelvic brace was measured, as shown in the picture below.

Figure 44: Actuator shaft coupled to the body via pelvic brace, with load cell reading the interaction forces.

The force signal was acquired at 10 kHz in LabVIEW, filtered and down-sampled to 200Hz. Open loop performance was investigated first, and can be seen in the figure below.
Figure 45: Servo amplifier’s inner current loop does not compensate for external conditions, like friction, which is likely responsible for the 15-20% steady state error.

The Xenus servo amplifier contains a current feedback loop, which aims to maintain the prescribed motor current draw, and therefore the actuator’s force output. The following control law from equation (3.3) was implemented:

\[ F_{\text{act}} = G(F_{\text{ref}} - F_{\text{ext}}) \]

Here \( F_{\text{ext}} \) is the force measurement from the load cell. Tests were performed with a proportional term only, with gains between 0.6 and 1.8. The step response with the gain set to 1.0 is in the figure below:
Figure 46: Closed loop step response with proportional gain $G=1$.

Here, the steady state error is approx. 50% and is a direct result of the control law and the low gain value used. The test was repeated with gains of 1.2, 1.4, 1.6 and finally 1.8, when instability occurred.

Figure 47: With a proportional gain of 1.8, serious instability occurred.
The step response test revealed serious instability with a proportional gain of 1.8, but even at lower gains, the subject could perceive vibrations in the system. These vibrations appeared above the proportional gain of 1.0. This gain value is not arbitrary, as it has been shown in [28] that a proportional force loop gain of 1 or larger can cause instability. Since such vibrations could have an adverse effect on the subject, it was decided that the force proportional gain of 1.0 represents a good compromise between safety/comfort and performance. As was shown in the previous chapter, a proportional gain used with our control law has the effect of reducing apparent inertia. Force feedback also increases backdrivability by reducing static and Coulomb friction.

### 5.2.3 Results of Force Loop Tests

As can be seen from the step input graph, gain $G=1$ produces a steady state error of 50% in the output with respect to the reference input. That result alone is unacceptable. Fortunately, this effect is compensated for by the impedance control algorithm. The equation which was derived earlier (3.3) for force command sent to the actuator is repeated here:

$$F_{act} = (G+1)[K_c(x_0 - x) + B_c(\dot{x}_0 - x)] + F_{ext}(G)$$

It can be seen that what is normally the reference force presented to the force loop, $F_{ref}$, is being pre-multiplied by the factor $(G+1)$. By introducing this factor, the reference force is offset to ensure that the force commanded to the actuator matches the force specified by the impedance gains ($F_{vir}$). Therefore, the steady state and tracking error due to the control law and low proportional gain is eliminated.
5.3 Impedance Controller - Bench Testing

5.3.1 Introduction

After tuning the force loop, another set of tests was performed to characterize the end point impedance controller. The main purpose here was to characterize the impedance controlled system and find the operating envelope. In our application, the core function of the impedance controller is to coax the actuator/robot end-point to display virtual stiffness and damping. In reality, as we have seen, the actuator also has inertia. Therefore, these tests also revealed the effects of the apparent inertia of the actuator, along with any friction and noise inherent in the system while it attempted to display the aforementioned virtual spring and damper qualities.

5.3.2 Testing Protocol

The performance of an impedance – controlled robot/actuator when interacting with an environment depends largely on the dynamic properties of that environment. Therefore, it was necessary for the actuator to interact with a human subject through the pelvic brace, in order to characterize its behavior [25]. The actuator’s end-point was attached to the pelvic brace worn by a subject just like in the force tuning experiment in the previous section. There are three possible inputs into our impedance controller: reference trajectory (position), measured position and measured force. Since it was difficult for the subject to produce repeatable position displacement, the subject was instructed to keep his body still while sinusoidal reference trajectories of different frequencies were presented to the controller.
One goal of the test was to find how well the actuator displays the commanded dynamic behavior. Therefore, the system’s load cell force data ($F_{\text{ext}}$) was acquired and compared against the force commanded by the impedance controller ($F_{\text{virt}}$). All tests were conducted with the force loop proportional gain $G=1$. The physical setup from the force loop tuning tests was used again, with the same subject for consistency.

Figure 48: The tests aimed to compare the forces resulting from the stiffness and damping constants ($F_{\text{virt}}$) and the external forces measured by the load cell ($F_{\text{ext}}$).

For the first set of tests, a single full sinusoidal cycle (1Hz, 3cm amplitude) comprised of 10,000 discrete points was continuously played back as a reference trajectory. The real-time control loop operated at 10 kHz, and the acquired data was down-sampled to 500Hz. For convenience, the derivative gains were specified through a selection of damping ratio
(zeta $\zeta$). The equivalent derivative constant $B_c$ was calculated from the knowledge of actuator’s moving mass, specified stiffness $K_c$ and desired damping ratio.

First the characteristic equation of a PD controlled actuator closed loop system was found. The actuator itself was represented as a second order system. The characteristic equation is:

$$s^2 + \left(\frac{B_c}{m} + \frac{b}{m}\right)s + \frac{K_c}{m} = 0$$

(5.1)

And the standard representation of second order characteristic equation is:

$$s^2 + 2\zeta\omega_n s + \omega_n^2 = 0$$

(5.2)

One can compare the two equations above and arrive at the following result:

$$B_c = 2\zeta \sqrt{m \cdot K_c}$$

(5.3)

For the first test, a relatively low stiffness value of 1kN/m was used, with damping values varied between 0 and 0.8.
Figure 49: Comparison of “virtual force” (spring only) versus the external measured force, acting on position error.

The figure above confirmed that in the general sense the impedance controller generated a proper magnitude force field. As the position error increased, so did the forces commanded by the impedance controller. The actual interaction force between the brace and the actuator somewhat closely followed the virtual force $F_{\text{virt}}$, with some oscillatory behavior. This can be explained by stiction in the bearings and lack of virtual damping in this particular test (aside from Coulomb friction inherent to the two linear guide bearings), which resulted in slight overshoot of corrective action. The offset between the virtual force $F_{\text{virt}}$ and the measured force $F_{\text{ext}}$ is due to the offset between the desired position and the actual position.
Figure 50: Comparison of “virtual force” (spring and damper) and the external measured force. The addition of damping practically eliminated the oscillatory force interaction. Irregularities in the otherwise sinusoidal shape of $F_{\text{ext}}$ just past maxima and minima were most likely caused by stiction.

With the damping ratio increased to 0.8, the measured force oscillations were eliminated.

The magnitude of measured force $F_{\text{ext}}$ still closely matched the virtual force commanded by the controller. Now that the system was shown to operate properly at low gains, tests were conducted at a higher virtual spring constant of 5kN/m with various frequencies and damping ratios. Some of the results are shown below.
Figure 51: At a higher virtual stiffness setting, the oscillations in the force reading are still apparent.

Increasing the virtual spring value produced an expected result: the actuator system produced greater environment deflections by exerting larger forces. Due to the greater force amplitudes, relatively speaking the effect of stiction is reduced compared to the 1kN/m test.
Figure 52: Once again, after the damping ratio (zeta) is introduced, the oscillatory behavior diminishes. As the damping ratio was increased during the test, the higher-frequency oscillations in position and force diminished. The derivative term was responsible for that, improving transient response characteristics.

Figure 53: With the reference trajectory frequency increased to 6Hz, actuator thrust rod’s inertia caused significant distortions to the position and force profiles. The system still behaved in a stable manner.
In Figure 53, it can be seen that actual position was asymmetrical. This effect was not seen before, even with the same virtual spring setting of 5kN/m. Since the test was conducted at 6Hz, which resulted in higher velocities, it can be stipulated that the damping magnitudes inherent to the environment were different for motion above the neutral position versus the motion below the neutral position, creating the asymmetry.

Figure 54: Introduction of damping had the effect of correcting the profile of the external measured force $F_{\text{ext}}$, by properly modulating the virtual force $F_{\text{virt}}$.

Adding a significant amount of damping to the controller produced a much more symmetrical displacement of the environment. It can be speculated that the virtual damping, being of relatively large magnitude, masked the damping inherent to the environment (pelvic brace). The anticipatory nature of the derivative term produced the commanded force signal $F_{\text{virt}}$, which had a phase lead with respect to desired position (as did actual position), and as a result the measured error $F_{\text{ext}}$ exhibited only slight phase lag with respect to position error.
Next, the stiffness was increased to 10kN/m. Some of the test results are presented below.

Figure 55: At this high virtual stiffness setting, slight vibrations were again felt, and can be seen in the $F_{ext}$ signal.

At a relatively high stiffness setting of 10kN/m, the controller behaved well, and the output forces closely matched the controller-commanded forces, with slight oscillatory behavior, which can be again attributed to lack of damping, as we have seen before. The behavior at higher damping ratios was investigated next.
Figure 56: Again, with the damping ratio increased, the vibrations diminish.

Figure 57: As the damping ratio was increased to 0.6, undesirable behavior appeared. The virtual damper component of the command signal began displaying vibratory behavior. The resulting forces were felt by the subject, but are not present in the measured force signal $F_{\text{ext}}$ due to low-pass filtering.
Figure 58: Increasing the damping ratio to 0.8 amplified the vibrations.

With the damping ratio increased past 0.6, the controller’s output signal began displaying high-frequency oscillations. They were perceivable to the subject, but due to low-pass filtering, these vibrations are not visible in the $F_{\text{ext}}$ signal. The most likely explanation is that the force loop – generated vibrations cause oscillations in displacement. Then, differentiation of position signal to obtain velocity amplified these oscillations, and with an increased damping ratio, their effect on the total signal is magnified.

In order to get a better understanding of the vibrations generated by the system at higher damping values, a power spectrum density graph of a sample data was generated. The figure below shows significant signal components at about 60Hz and 500Hz.
Figure 59: Unfiltered position feedback contains significant noise at about 60Hz and 500Hz. 

A solution to the vibration problem was proposed. An attempt was made to filter the derivative portion of controller output. Unfortunately, the time delay which the digital filters introduced actually deteriorated the system’s behavior, and the vibrations increased in magnitude. The same effect has been described in literature [29]. Therefore, a decision was made to establish a limit on the damping ratio and position and force proportional gain, below which these undesirable vibrations do not occur.

The system was then tested at 10kN/m virtual stiffness and 3Hz reference trajectory. The results are presented below.
Figure 60: With the reference trajectory of 3Hz and no damping, the measured force signal ($F_{ext}$) tends to lag behind the commanded force ($F_{virt}$).

Figure 61: Increase in damping ratio smoothed out both force curves.
Figure 62: With the damping ratio set to 0.5, the system still behaved well.

Figure 63: Once the damping ratio was increased to 0.6, the performance deteriorated due to appearance of high frequency vibrations, which can be seen in the $F_{virt}$ signal, and could be felt by the subject.

The same testing procedure was used to investigate system’s behavior at 6Hz, which is the upper extreme of the frequency range of interest. Due to higher velocities and
accelerations, the inertial effects of the actuator thrust rod mass were expected to be the greatest here.

Figure 64: Inertial effects cause the measured force profile $F_{\text{ext}}$ to lag significantly behind position error.

Figure 65: Increasing the damping ratio seemed to make the controller efforts ($F_{\text{virt}}$) more abrupt.
Figure 66: As seen before, damping ratio of 0.6 amplifies the derivative action of the PD impedance controller. This causes high frequency vibrations.

The last two graphs confirmed findings from previous tests. Increasing the damping ratio and therefore adding the derivative term tends to reduce phase lag of the measured force $F_{ext}$ compared to the desired position. Once the damping ratio was increased past 0.5, an undesirable behavior occurred.

### 5.3.3 Impedance Test Results

In light of the results from the preceding tests, the operating envelope of the system was defined by limiting the maximum allowable damping ratio to $\zeta=0.5$, and the virtual stiffness to $10\text{kN/m}$. Using equation (5.3) with $K_c=10\text{kN/m}$, $m=1.9\text{kg}$ and $\zeta=0.5$, the maximum allowable virtual damping is $138\text{N/m/s}$.
Though the system response was tested at damping ratios higher than 0.5, it was unclear what the range of desired damping may be when applying corrective force fields to the pelvic area. It can be argued that it is the virtual spring stiffness, which has the greatest bearing on effectiveness of robotic gait rehabilitation, and a certain minimum amount of damping could be introduced into the system purely for the purpose of obtaining better system behavior, rather than getting certain first order endpoint impedance. The set of tests conducted made it possible to move forward with testing the system on healthy human subjects, in order to get a better insight into the range and type of endpoint impedances necessary to facilitate rehabilitation.

5.4 Selection of Desirable Impedance Gains

5.4.1 Introduction

The body’s soft tissue and the pelvic brace exhibit a certain amount of admittance (inverse of impedance). From the point of view of a subject’s skeleton, where the subject wears a brace the movement of which is restricted by an outside structure, the body’s soft tissue and the brace impede on the movement of the skeleton, the dynamics of which will be approximated here by a 1st order system.

5.4.2 Actuator – Body Dynamics

The task of an end – point impedance controller is to coax the actuator/robot end point to exhibit the desired impedance. This generated virtual stiffness and damping, combined with stiffness and damping of the brace, results in the total impedance applied to the body
(Figure 67). This total impedance will be called here the desired impedance, comprised of desired stiffness and desired damping.

![Diagram](image1)

**Figure 67:** Virtual stiffness and damping applied to the body.

Due to the fact that in most practical applications it is impossible to completely eliminate the effects of the actuator’s mass, the illustration below comes closer to describing the true dynamics of actuator – body interaction.

![Diagram](image2)

**Figure 68:** In addition to virtual stiffness and damping applied to the body, there also exists apparent inertia of the actuator.

In general it is advantageous to diminish the effect of the apparent inertia of actuator mass $m_{\text{act}}$ and that is achieved with force feedback, but often it cannot be less than 50%
of actual mass. One approach to dealing with this apparent inertia is simply accepting its existence, and equating it to the desired inertia \( M_d \). From here on, the remainder of impedance which we are interested in applying onto the body is that consisting of stiffness \( K_d \) and damping \( B_d \). The total (desired) “stiffness” experienced by the body is the reciprocal of the sum of the reciprocals of the actuator’s stiffness and the pelvic brace stiffness:

\[
\frac{1}{K_d} = \frac{1}{K_c} + \frac{1}{K_e}
\]  
(5.4)

And therefore the stiffness to be exhibited by the actuator is

\[
K_e = \frac{K_d K_c}{K_e - K_d}
\]  
(5.5)

The total damping experienced by the body is:

\[
B_d = B_c + B_e
\]  
(5.6)

And the damping to be applied by the actuator is:

\[
B_c = B_d - B_e
\]  
(5.7)

### 5.4.3 Conclusion

Since there are quite large variations in body types in the general population, the body admittances exhibited by different subjects are expected to vary rather greatly. In addition, any variations in the way a pelvic brace is worn as well as any other changes made to the brace interface points can change its inherent mechanical properties. As we saw in the previous section, the total impedance (and therefore force field) experienced by a subject is the sum of the impedance inherent to the body/pelvic brace, and that exhibited by the actuator/mechanism. For the purpose of comparing results of therapy in
the future, and for other purposes, knowledge of the total force fields experienced by the subject is necessary. Therefore, a procedure for measuring subject’s and pelvic brace’s dynamics is necessary, and then using the relationships from equations (5.5) and (5.7), the controller gains can be automatically selected based on the desired total impedance.

The equations describing the relationship between the robot’s end point impedance with that of the environment produce interesting results. They suggest that the maximum total stiffness applied onto the body cannot be greater than either the robot’s end-point virtual stiffness, or the environment’s inherent stiffness. Therefore, it may be advantageous to design the haptic interface such that the combined environment impedance is maximized. This would maximize the adjustment range of the total impedance. On the other hand, an environment displaying higher impedance, and more specifically higher stiffness, can have a detrimental effect on the stability of an impedance – controlled system [25]. In that case, the necessary force loop gain reduction (to maintain stability) would increase the apparent inertia of the system (as experienced by the environment – the subject).

Now, going along with the assumption stated earlier, that large apparent inertia of the robot is undesirable and should be minimized, it would mean that excessive environment stiffness would result in substituting a greater impedance adjustment range for that impedance’s quality. In conclusion, an optimum stiffness of the haptic interface (the pelvic brace) will be found in the future, such that operating range and system’s performance are optimized.
Chapter 6. Moment – Fields Applied to Pelvic Obliquity with Impedance Control

6.1 Introduction

In the previous chapter the behavior of our end – point impedance controlled linear actuator system was characterized. The system interacted with a subject’s pelvic area through the Newport 4 pelvic brace and the controller was implemented at the actuator’s linear motion level. In this chapter, we describe how the system was adopted to apply moment – fields at the obliquity level of the pelvis.

6.2 Obliquity Measurement

Application of corrective torques/forces to the pelvic area to impede hip-hiking requires measurement of obliquity at any given instant. In the field of human motion analysis, pelvic obliquity is specified in degrees of angular rotation. In order to comply with this standard, we decided to offer position feedback to the controller in the same format. The main constraint imposed by the design of the RGR Trainer II when it came to pelvic obliquity measurement was the use of two linear actuators, which are attached to either side of a pelvic brace and operate in the vertical direction. The STA2504 servo tube actuators have built-in position sensors. Therefore, it was decided to calculate obliquity based on the relative position of the two attachment points on the pelvic brace and the distance between these two points. This is illustrated in Figure 69.
Figure 69: Obliquity angle can be calculated knowing vertical position of the two attachment points. $D$ is the length of the direct line between the two attachment points, and $y$ is the distance between them in the vertical direction. Due to deformations from loads, the measured obliquity $\Theta$ differs from pelvis’ true obliquity angle.

As one side of the pelvis moves upward with respect to the other, the segment of length $D$ spanning the two attachment points rotates. The length of this segment depends on the subject’s body size and the size of the brace, and can be easily measured. If this segment “$D$” serves as the hypotenuse of a right triangle and the difference in height between the two attachment points is “$y$”, then the resulting angle of rotation “$\Theta$”, which can be found with trigonometry, is the obliquity angle.

Of course, due to the flexibility of the haptic interface and the body’s soft tissue, the angular rotation as measured at the pelvic brace is not equal to the true obliquity angle of the pelvis. In fact, only when there are no moments/forces applied to the brace can this measured obliquity angle $\Theta$ be equal to the true obliquity angle of the pelvis. Therefore, from here on, the angle $\Theta$ will be referred to as “measured obliquity”. The result is such
that this system controls end-point impedance at the measured obliquity level, and not the impedance at the true pelvic obliquity level.

The RGR Trainer II was originally designed with two linear actuators, one on each side, but only one actuator is required to target hip-hiking on one side of the body. Since only one Xenus servo amplifier was available at the time, this task was accomplished with just one actuator. In order to obtain measured obliquity, a linear potentiometer was used on the opposite side of the pelvis for position measurement. A light assembly was built, which provided guidance for the linear potentiometer, as seen in Figure 70.

![Figure 70: Picture of the pelvic brace with thigh segment, servo-tube actuator and linear potentiometer.](image-url)
6.3 Application of Moment – Field to the Pelvis

During gait, as one leg goes through the swing phase, the opposite leg is in stance. The hip joint of the stance leg serves as the pivot point of the pelvis as it undergoes obliquity. Therefore, the distance between the actuator attachment point and the stance hip joint (the pivot point) is the moment arm “d”. The length of the moment arm can be found knowing the distance between attachment points “D” and the distance between the hip joints “c”, as shown in Figure 71 below.

![Figure 71: A linear actuator applies a corrective torque to obliquity through a moment arm of length “d”.

In order to apply impedance control at the obliquity level, the control algorithm from the linear — motion case was adapted to act on position error in degrees of measured obliquity. The system diagram is presented in Figure 72. The PD gains consist of the proportional gain $K_c$ specified in terms of desired moment command as a response to
angular position error, and the derivative term operating at the actuator’s linear motion level and specified in terms of force command as a response to velocity error, as seen in Figure 73. For the purpose of transforming rotational velocity to linear velocity, a linear approximation to the Jacobian was used. With the motion limited to about 20 degrees, this approximation’s deviation from the actual Jacobian transformation is minimal.

Figure 72: The PD controller acts on the obliquity error and outputs a proper force command.

Figure 73: The proportional gain $K_c$ is specified at the obliquity level, while the derivative gain $B_c$ acts on linear velocity error at the actuator level.
As was explained in the previous chapter, for convenience, the derivative gain $B_c$ was not specified independently, but was computed by specifying the desired damping ratio at the actuator level. Therefore, the obliquity level proportional gain $K_c$ is converted to its linear equivalent, and equation (5.3) is used to find the derivative term $B_c$:

$$B_c = 2\zeta \sqrt{m*K_c}$$

This type of approach to gain selection allows fast changes to be made to the moment-field strength while the general dynamic properties of the system remain unchanged.

### 6.4 Collection of Obliquity Reference Trajectories

#### 6.4.1 Introduction

In order to facilitate robotic gait retraining targeting pelvic obliquity using our control scheme, realistic healthy—subject motion trajectories of pelvic obliquity had to be provided to the controller as a reference input. With help from Dr. Benjamin Patritti from the Motion Analysis Lab at Spaulding Rehabilitation Hospital in Boston, Massachusetts, lower extremity motion profiles of a healthy 30 year old subject were collected using Vicon motion capture system.

#### 6.4.2 Test Protocol

The Vicon system used reflective markers, which were placed on the legs. Four markers were also placed on the right and left anterior superior iliac spines (LASI and RASI) and posterior superior iliac spines (LPSI and RPSI), as shown in Figure 74.
The subject ambulated on treadmill for several minutes until comfortable. The subject’s comfortable walking speed (CWS) was found to be 3km/h. Since the RGR Trainer II was designed with a grab-bar for safety and support, and since it was anticipated that pelvic obliquity might vary depending on whether a grab-bar is used by the subject or not, two
types of data were collected. A horizontal grab-bar was rigidly mounted in front of the subject and at 60% of his height in order to investigate the “hold” vs. “no hold” condition on pelvic obliquity. Motion data was acquired at 15 different speeds ranging from 1 km/h to 5 km/h. Two 30-second samples were collected per each walking speed: one for hold-condition and one for no-hold condition. The full list of speeds collected is presented in the table below.

<table>
<thead>
<tr>
<th>km/hr</th>
<th>m/s</th>
<th>% of CWS</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>1.00</td>
<td>0.33</td>
</tr>
<tr>
<td>2</td>
<td>1.20</td>
<td>0.40</td>
</tr>
<tr>
<td>3</td>
<td>1.40</td>
<td>0.47</td>
</tr>
<tr>
<td>4</td>
<td>1.80</td>
<td>0.60</td>
</tr>
<tr>
<td>5</td>
<td>2.20</td>
<td>0.73</td>
</tr>
<tr>
<td>6</td>
<td>2.40</td>
<td>0.80</td>
</tr>
<tr>
<td>7</td>
<td>2.60</td>
<td>0.87</td>
</tr>
<tr>
<td>8</td>
<td>3.00</td>
<td>1.00</td>
</tr>
<tr>
<td>9</td>
<td>3.40</td>
<td>1.13</td>
</tr>
<tr>
<td>10</td>
<td>3.60</td>
<td>1.20</td>
</tr>
<tr>
<td>11</td>
<td>3.80</td>
<td>1.27</td>
</tr>
<tr>
<td>12</td>
<td>4.20</td>
<td>1.40</td>
</tr>
<tr>
<td>13</td>
<td>4.60</td>
<td>1.53</td>
</tr>
<tr>
<td>14</td>
<td>4.80</td>
<td>1.60</td>
</tr>
<tr>
<td>15</td>
<td>5.00</td>
<td>1.67</td>
</tr>
</tbody>
</table>

Mary Goldsmith, a graduate student at Boston University and a Research Assistant at Motion Analysis Lab created a custom Matlab program to extract pelvic obliquity from
the 3-D trajectories of four reflective markers on the pelvis, while the markers on the legs were used to find the percentage of gait cycle. Since each 30-second trial contained several full gait cycles, the Matlab program found a representative gait cycle per each walking speed by averaging 15 full gait cycles. Sample data and the result of averaging are shown in Figure 76 and Figure 77.

![Pelvic Obliquity](image)

Figure 76: Sample obliquity angle data for 15 cycles, collected at the same walking speed (here 1.4 km/h).
6.4.3 Obliquity Capture Results

Collection of motion data with the camera based Vicon system, and extraction of trajectories of interest by Mary Goldsmith gave us access to realistic pelvic obliquity data at various speeds. Having average obliquity cycles allowed for healthy subject testing of obliquity – level impedance control, while detailed analysis of the data for effects of hold vs. no hold and other conditions could be done in the future.

Figure 77: Sample averaged trajectory of data from Figure 76.
6.5  RGR Trainer II – Obliquity Level Tests

The following testing of a healthy subject was conducted under an approved and active Institutional Review Board (IRB) application at Spaulding Rehabilitation Hospital in Boston, Massachusetts.

6.5.1  Backdrivability Test - Introduction

In Chapter 5, a series of bench tests were conducted to get an insight into the characteristics of our impedance controller at actuator level. The subject remained still, while sinusoidal reference trajectories were presented to the controller. There was no attempt made to quantify the backdrivability of the system, since there was no practical way to present to the controller through the feedback path a realistic pelvic obliquity signal. Once the system was adapted to apply moment – fields at the pelvic obliquity level and we had access to realistic pelvic obliquity data, it became possible to test the system’s backdrivability and its ability to properly apply corrective moment – fields in response to hip – hiking simulated by a healthy – subject.

6.5.2  Backdrivability Test Protocol

A 30 year old, 80kg and 1.80m tall healthy male subject donned the Newport 4 pelvic brace with left and right thigh components and entered the device (Figure 78). The linear actuator and linear potentiometer end points were attached to the brace on either side of the body using spherical joints (Figure 79).
Figure 78: The subject within the RGR Trainer II, standing on treadmill.

Figure 79: Close-up of actuator and potentiometer end points interfacing with the Newport 4 pelvic brace.
A reference trajectory of pelvic obliquity (no holding) collected at 1.4 km/h had been up–sampled from Vicon’s 120 Hz capture rate to the controller’s 10 kHz loop rate. A LabVIEW program designed to apply moment–fields via obliquity–level impedance control was used to conduct the test. The program’s GUI presented data in real time to both the system operator and the subject. Waveform graphs displayed the desired obliquity $\Theta_d$ (reference trajectory), measured obliquity $\Theta_{\text{meas}}$, virtual moment $M_{\text{virt}}$ commanded by the impedance controller and measured moment $M_{\text{ext}}$. With all gains set to zero (including the commanded virtual stiffness and damping values) the program was launched. Launching the program began recording of position and force signals (measured and desired) as well as the values of gains used. With the GUI clearly visible to the subject, the treadmill was brought up to speed of 1.4 km/h (minimum adjustment size was 0.2 km/h). Once the subject synchronized his gait cadence to the reference trajectory and attempted to maintain proper obliquity magnitude, the force loop’s proportional gain $G$ was slowly increased by 0.1 increments, such that several gait cycles under each gain setting were obtained. A gradual reduction of interaction forces was observed as the gain was increased.

### 6.5.3 Backdrivability Test Results

With the gain $G=0$, the interaction forces ranged between -10N and +10N, as can be seen in Figure 80. There exist points in the data where the recorded forces are -15N and more. These peaks seem to coincide with left heel strikes. This makes sense. The actuator was
attached the pelvic brace on the left side, and a heel strike during gait would have caused rapid deceleration of the thrust rod, resulting in significant inertial forces.

![Backdrivability, G=0.0](image)

Figure 80: Forces measured by load cell generally ranged between -10 and +10 N, with peaks over 15N at heel-strike and toe-off.

At the other extreme, with the gain G=1, the forces experienced by the subject were reduced to roughly between -5N and +5N. The peaks in force coinciding with heel strikes seem to be unaffected, maintaining their original amplitudes, as seen in Figure 81. It is likely that due to its limited bandwidth, the actuator could not respond to control inputs fast enough to attenuate these force peaks. Outside of the heel strike force peaks, the result of implementing force feedback to increase backdrivability was as expected. By using a proportional gain G=1, the measured forces were reduced by about 50%.
Figure 81: Though the heel-strike and toe-off peaks remain, the interaction forces for the remainder of the gait cycle are reduced by 50% with a gain of G=1.0.

Concerning the subject’s comfort and fit of the pelvic brace, the subject reported that while the pelvic brace felt comfortable, the thigh components limited his range of motion. The inner portions of the thigh components created interference at the knee level and necessitated the subject to modify his gait by abducting his legs to create clearance when in the swing phase. Also, since the center of rotation of the joint at the hip-level connecting the pelvic component to the thigh component was not coincident with the hip joint itself, the abduction was largely restricted and caused the brace to migrate.
6.5.4 Environment Dynamics Measurement – Introduction

In the preceding tests, the behavior of the end-point impedance controller was investigated. There was no need to consider the environment dynamics except for its effect on force feedback. We saw in Chapter 5 that the total force-fields experienced by a subject depend on both the robot end point impedance and environment dynamics. Therefore, in order to investigate the effect of force field on pelvic obliquity, it became necessary to measure the environment dynamics. More specifically, the emphasis was placed on environment stiffness. That’s because stiffness is responsible for the bulk of forces applied onto the body. In the future, tests investigating effects of damping on subjects displaying real secondary gait deviations could be performed.
6.5.5 Environment Dynamics Measurement - Protocol

The subject wore the Newport 4 pelvic brace with thigh components and entered the RGR Trainer II just as in the backdrivability test. A LabVIEW program running at 10 kHz generated sinusoidal force commands at a frequency of 4Hz, which the Xenus amplifier executed using its internal current feedback loop when driving the servo tube actuator. Force feedback was not used. Three force amplitudes were used: 40N, 60N and 80N. The said LabVIEW program computed the environment stiffness value $K_e$ in real time. An algorithm found the displacement’s maxima and minima (and therefore the amplitude) and the corresponding force readings at that instant, and computed the environment’s stiffness $K_e$. The result was available at the actuator level (N/m) and at the obliquity level (N-m/deg).
6.5.6 Environment Dynamics Measurement – Results

A sample data from the test acquired at 100Hz is shown in the figure below.

![Stiffness Measurement - 80N](image)

Figure 83: Sample data from stiffness measurement test.

The results of stiffness measurement at the three force amplitudes are summarized in the table below.

<table>
<thead>
<tr>
<th>Force (N)</th>
<th>$K_e$ [N/m]</th>
<th>$K_e$ [N-m/deg]</th>
</tr>
</thead>
<tbody>
<tr>
<td>40 N</td>
<td>5000</td>
<td>12.5</td>
</tr>
<tr>
<td>60 N</td>
<td>4200</td>
<td>10.5</td>
</tr>
<tr>
<td>80 N</td>
<td>3900</td>
<td>9.75</td>
</tr>
</tbody>
</table>

Table 3: Environment stiffness measurement results.

The measurement procedure produced relatively good estimates of the environment stiffness $K_e$, as suggested by the fact that doubling the force amplitude from 40N to 80N
resulted in a stiffness change by only about 20%. Of course, this method can be improved on in the future to become a standard part of device calibration performed at the beginning of each training session.

An interesting effect was found during the test when the subject tightened up his thigh muscles. An immediate and significant increase in stiffness measurement was seen. This result makes sense, since the brace assembly relied greatly on the thigh components for “adhesion” to the pelvic area. This result also served as a confirmation that the method is working as expected.

6.5.7 Introduction to Moment – Field Application Test

One of the main questions, which waited to be answered until now, was how a force – field of a certain strength applied at the obliquity level feels when a subject deviates from the prescribed trajectory. An end point impedance controller had been adapted to operate at the obliquity level, and now that the environment admittance has been measured, it became possible to answer that question. As we have seen in section 5.4.2, the total stiffness experienced by the subject, called here the desired stiffness $K_d$, can be found from equation 5.4:

$$\frac{1}{K_d} = \frac{1}{K_e} + \frac{1}{K_v}$$

This idea was put in practice by using the subject’s environment stiffness and tuning the actuator’s end point stiffness to define the total stiffness experienced by the subject.
6.5.8 Moment – Field Application Test - Protocol

An end point obliquity-level impedance controller was implemented in LabVIEW as described in section 6.3, with the following selectable settings in the user interface: attachment span “D”, hip joint span “c”, environment stiffness $K_e$, desired stiffness $K_d$, damping ratio and control command adjustment. The controller featured a computation of the virtual stiffness $K_c$ from a desired stiffness $K_d$ and environment stiffness $K_e$, as well as a computation of the moment arm $d$ from knowledge of attachment span $D$ and hip joint spacing $c$. For our subject, $c = 0.18m$ and $D=0.56m$, resulting in a moment arm $d=0.37m$. A screenshot of the user interface is shown below.

Figure 84: The graphical user interface contained real-time displacement and force information, as well as control parameter inputs.
The subject entered the RGR Trainer II while wearing the Newport 4 pelvic brace, and the left and right sides were attached to the actuator end point and the potentiometer end point respectively, as was done in the backdrivability test. In the user interface, an environment stiffness of 12 N-m/deg was entered into the proper field and a damping ratio of 0.3 and force proportional gain G=1.0 were selected.

Upon starting the control program, a reference trajectory “no-hold 1.4kmh” was launched. While this trajectory was being continuously looped and presented to the controller as the position reference, it was also displayed on a screen visible to the subject. The treadmill was slowly brought up to the speed of 1.4 km/h, and the subject synchronized his gait to the reference trajectory. Once comfortable, the subject began simulating hip-hiking, by raising his left hip as the left leg was in the swing phase. As the total desired stiffness $K_d$ was gradually increased up to 6 N-m/deg, that setting in combination with the environment stiffness input $K_e$=12 N-m/deg produced a proportional controller gain $K_c$=12 N-m/deg. Since the system did not feature a direct reading of true pelvic obliquity, the impedance controller acted upon actuator/potentiometer end point obliquity, referred-to here as “measured obliquity”, and not the true obliquity of the pelvis. This effect had been accounted for by first measuring the environment’s stiffness, and using it to calculate the necessary actuator end-point stiffness, in order to produce the total desired effect.
6.5.9 Obliquity – Level Test Results

The highest selected “desired stiffness” $K_d$ value of 6 N-m/deg resulted in the controller proportional gain $K_c$ of 12 N-m/deg, which acted upon the end-point “measured obliquity”, results of which can be seen in Figure 85.

![Figure 85: System’s response elicited by simulating hip-hiking at 1.4 km/h. “Measured Obliquity” is generally less than the subject’s true pelvic obliquity.](image)

The peaks in “measured obliquity” of about 7 degrees resulted from the subject raising his left side of the pelvis and simulating hip-hiking while the left leg was in swing – phase. In the remaining portions of the gait cycle, the subject’s obliquity matched the reference trajectory rather closely. In response to the position and velocity errors from the simulated hip-hiking, the PD gains produced a control effort, which is shown here as a virtual moment applied to the subject’s pelvic area, $M_{virt}$. As the moments applied to the body peaked above 60 N-m, the corresponding forces produced by the actuator were...
on the order of over 160N. In order to produce such displacement errors, the subject exerted considerable amounts of energy. The subject reported that the force field’s high strength helped synchronize his gait cadence to the reference trajectory’s cadence. The haptic feedback provided to the pelvis alerted the subject of any discrepancies in the cadence, resulting in a correction.

As the subject produced even higher position errors, the Xenus servo amplifier’s thermal protection responded by saturating the control input to the steady-state value of 51.2N (for servo tube actuator model STA2504). This means that in order to exert moments greater than 160 N-m cyclically with similar frequency, the moment-arm length should be increased appropriately, or a more powerful actuator should be used.
Chapter 7. Conclusions

7.1 Results from Impedance Control Implementation in RGR Trainer II

The second prototype of the Robotic Gait Rehabilitation Trainer served as a platform for the implementation of impedance control, in order to apply corrective force-fields to the pelvic area for gait retraining. Despite the fact that the mechanism design was not ideal, the decision was made to focus on control algorithm development and implementation, while only the changes necessary to achieve the main goal would be implemented. Therefore, despite the fact that the mechanism’s friction forces were found to be unacceptably high, limiting free motion in the device, the issue was not addressed because implementation and testing of force field application to the pelvis could still be achieved. No attempt was made to use the weight – support system originally designed by the undergraduate team for the same reasons.

Human testing revealed, that the design layout did not allow for arms to swing freely. For safety reasons, there is merit to providing grab-bars in front of, or on both sides of the patient. Pelvic motion data for both “hold” and “no-hold” conditions were acquired for future analysis, so that a respective trajectory is available for either condition. Nevertheless, an attempt will be made to design the next RGR Trainer such that the subject can swing arms freely and to give physical therapists an unrestricted access to lower extremities.
The impedance control algorithm was implemented in the RGR Trainer systematically, by first implementing it at the actuator level. Once the controller operated well at the actuator level, it was bench tested while interacting with a compliant environment – a human subject wearing an orthotic pelvic brace is. Bench testing under those conditions allowed us to define an operating envelope. Finally, the impedance controller was implemented at the pelvic obliquity level. Treadmill testing within the device demonstrated that the system works as expected. The backdrivability test presented solid proof that even without force feedback, the device imposes little impedance onto the subject (at the obliquity level), thanks to the highly backdrivable servo tube actuator. With force feedback, the interaction forces were reduced by 50%, a direct result of force control law used. Finally, the force-field/moment-field application program was tested. A healthy subject simulated hip-hiking while walking on treadmill, and in response the system applied corrective forces. This test proved that the idea of using impedance control to apply force fields to the pelvis works well. In addition, for the first time the torques required to affect pelvic obliquity were estimated.

### 7.2 Future Work

#### 7.2.1 Machine Design

The RGR Trainer will be redesigned based on the knowledge gained from testing the second prototype. We will aim to design a device and mechanism, which are largely
unobtrusive to the patient, with high intrinsic backdrivability. One embodiment of such a device is shown in Figure 86.

![Figure 86: Concept of RGR Trainer III.](image)

This design targets pelvic obliquity and pelvic rotation. The mechanism is designed to transmit rotational moments to the pelvic brace, as opposed to transmitting linear forces. Therefore, throughout the gait cycle, there are no variations in moment arm length, and no undue variations in the effective moments acting onto the body. The design lends itself well to either passive or active weight compensation via the same push-rods, which
control pelvic obliquity. Overall, the layout of the actuation mechanism, being located behind the subject and from the waist – level up, provides unrestricted access to the legs.

7.2.2 Impedance Controller

Due to changes in the actuation mechanism design and sensor location, force feedback design will be taken into account when designing the new mechanism. The general architecture of the impedance controller will not change. The system’s performance, especially with regard to the “force loop” is greatly influenced by force sensor location due to non-collocation issues. These problems arise due to the interaction between the mechanism’s structural modes and the force feedback loop [25]. In general, force feedback can reduce the apparent inertia as experienced by the environment (subject), but this pertains only to the mass located between the actuator and the load cell. On the other hand force feedback does not change the apparent inertia of the mass located between the environment and the force sensor. Therefore a compromise will be found between transmission link stiffness and its mass, versus force sensor location, in order to attain best performance.

7.2.3 Machine – Patient Synchronization

During human testing described in Chapter 6, the subject actively adjusted his gait cadence to match that of the reference trajectory. With some practice, this task was manageable, but active synchronization to the reference trajectory cannot be expected from a post-stroke patient. Therefore, a gait prediction algorithm will be developed. This system will likely operate as a state machine, which will use inertial measurement
units (IMU’s) and foot switches to gather data from the subject and predict both the location in the gait cycle and the cadence. This information will be used to adjust the reference trajectory, by scaling it in time, therefore minimizing gait location discrepancy between the subject and the device.

7.2.4 Haptic Interface Design

Human testing of the device also revealed a number of problems with the current haptic interface – the Newport 4 pelvic brace system. Since this orthosis was designed to restrict motion in post-operation hip revision patients, it restricts movement in the frontal plane: the thigh components cause interference at the knee joint and the non-coincident centers of rotation (with the hip) restrict thigh abduction. Therefore, these effects will be investigated more closely, and if necessary, custom gear will be designed to improve the haptic interface.

7.2.5 Human Testing

Finally, human testing will be performed. Testing will be divided into three sets of experiments. The first set of experiments will be used to demonstrate that the gait prediction algorithm works properly. The Vicon camera motion capture system will be used to validate the results obtained by the gait prediction algorithm. During the second set of experiments we will investigate the response of post-stroke hemiplegic patients to the force-fields generated by the RGR Trainer. Three modes will be tested: pelvic obliquity control only, pelvic rotation control only, and pelvic obliquity and pelvic rotation combined. Finally, the last set of tests will combine haptic and visual feedbacks
for gait retraining in post-stroke patients. While the subjects receive haptic feedback regarding their pelvis, corresponding to the secondary gait deviations, they do not receive any feedback regarding their primary gait deviation location – the knee joint. Therefore, the knee flexion angle will be continuously measured, and its value will be displayed to the subject in real time. A simple visual display will feature the measured knee flexure angle compared against a set point. As with the earlier set of tests, a motion capture system will be used to validate the results.
Appendix A. The SHIMMER Packaging Redesign

A.1 Introduction

SHIMMER (Sensing Health with Intelligence, Modularity, Mobility and Experimental Reusability) is a compact extensible wearable sensor package, originally designed by Intel. It features an integrated 3-axis accelerometer, large storage, low power requirements and wireless communication. With additional expansion boards (gyroscope board, EMG board and others), the SHIMMER can be used for motion sensing as well as ECG, EMG and EKG data collection and transmission. As designed and built by Intel, SHIMMER’s power is supplied by a rechargeable 3.7V - 250mAh lithium polymer battery. Depending on the mode of operation, this battery allows for between six and eight hours of operating time. The SHIMMER’s electronic hardware is encased in a sleek, custom-designed plastic enclosure.

Figure 87: SHIMMERs, shown full-size, with a gyro board on the left, and an EMG board on the right.
Figure 88: SHIMMERs plugged into the strip charger.

Dr. Paolo Bonato, the director of Motion Analysis Laboratory at Spaulding Rehabilitation Hospital in Boston, Massachusetts, and his associate, Shyamal Patel, were preparing to conduct a Parkinson’s disease study using SHIMMERs. From their previous experience with the units, they identified four key upgrades, which would greatly enhance SHIMMER’s usability in this study, considering that the subjects would wear the units while at home. These upgrades included housing a new larger battery, incorporating an on/off switch to save power, firmly locating the internal electronic components within the enclosure to avoid noise in data, and making a provision for a hook & loop type strap to affix the sensor package to the patient’s body. We were asked by Dr. Bonato to design a new enclosure for the SHIMMER, which would incorporate all of the improvements listed, while keeping overall size to a minimum and using the most economical manufacturing solution at the volume of sixty enclosures. The project would also include updating the charger body to fit the new design.
A.2 Background

The Motion Analysis Lab’s solution to short battery life in previous studies was to attach a larger, 500mAh battery to the outside of the SHIMMER with double sided tape. The battery was adopted from the popular Ipod Nano personal media device. This quick solution increased run time to a satisfactory level, but due to safety and other concerns, these units could not be used outside of the lab. A number of the original SHIMMERs were also outfitted with small toggle on/off switches, which were glued on the inside of the enclosure. The switch was activated by use of a paper clip, through a machined slot. This solution gave the staff more control over battery life, but it reduced the unit’s ease of use and practicality. The picture below shows different components of the original SHIMMER with the switch glued in place and an access slot in the side of the clear top enclosure.

Figure 89: SHIMMER’s original components. Top row from left to right: bottom enclosure piece with switch glued in place, clear top enclosure and EKG top enclosure. Bottom row: motherboard - printed circuit board (PCB) with Bluetooth module, original 250mAh lithium polymer battery with protection circuit module (PCM), kinematics three axis gyro expansion board and EKG/EMG/ECG expansion board.
A.3 Design Process

As a first step in the task of redesigning the enclosure, a proper battery had to be chosen. The size of the battery would ultimately have most influence on the size of the whole package. Benjamin Kuris, an electrical engineer at Intel familiar with the project, strongly suggested that a protection circuit module-equipped battery from the same company be used. A 580mAh battery from Ultralife Batteries was identified. Unfortunately, this battery was significantly larger than the 500mAh battery from an Ipod Mini, which had been used successfully at Motion Analysis Lab before. Initially in the design process, there was a probability that Intel would assist in manufacturing the units, in exchange for using all of the intellectual property contained in this design project for their own purposes. Since Intel would only accept a design which used the safer, larger 580mAh battery, it was decided to pursue two design concepts, using both batteries. The three batteries are pictured below. Also, a single pole double throw miniature slide switch from Philmore was selected to provide the power on/off function.

Figure 90: Size comparison between the original SHIMMER battery on the left and the two possible replacements.
The CAD work was done using SolidWorks 2007 software. In case of prototype 1, both versions of the enclosure consisted of four parts, namely the enclosure, the bottom cover and two top covers: top cover low and top cover high, to accept different size expansion boards.

The two parts of the original SHIMMER enclosure were held together with four snap-fit features and a single 2-56 machine screw and nut. This solution did not keep internal components from shifting around. One of the core functions of the SHIMMER is collecting movement data with its built in accelerometers and its gyroscopic sensors. The electronics were kept from gross shifting around, but fine movement was only restricted by friction. Therefore, the data collected was often indistinguishable from the noise. In case of the motherboard, this problem was solved in both design variants (500mAh and 580mAh) by using four screws in four corners of the enclosure to clamp the motherboard down in place. The pictures below show the two variants of the 500mAh design and the 580mAh design.

Figure 91: 500mAh version of prototype 1, with top-cover-low on the left and top-cover-high on the right.
Figure 92: 580mAh version of prototype 1, with top-cover-low on the left and top-cover-high on the right.

The interface between the two parts was designed with a 0.010” clearance, in order to make sure that contact is made on the motherboard first. The size of the clearance was not picked arbitrarily. It was expected from the beginning that the manufacturing method to produce 60 units would be one of the rapid prototyping ones. The Robotics and Mechatronics Lab at Northeastern University in Boston, Massachusetts, owns a stereo lithography apparatus (SLA) – Viper Si² from 3-D Systems. This is where pre-production proof-of-concept prototypes would be built. The SLA system builds solid models from CAD data by solidifying 0.002” thick layers of liquid resin, using laser, and successively forming the geometry from the bottom – up layer by layer. Due to the resolution limitations of the machine, and accounting for tolerance stack-up, a designed-in clearance of 0.010” would ensure that the minimum actual clearance in the enclosure would be 0.002”. This clearance would still result in the screw clamping force “sandwiching” the motherboard in place. On the other hand, greater clearance would only result in greater distortion in the plastic parts, and undue internal stresses. The
The pictures below show the CAD model of the 500mAh enclosure and a close up of the built-in clearance.

Figure 93: Dimensional limitations of Viper SLA system. With the above tolerance levels, the preload acting on the PCB is between 0.002” and 0.008”.

The battery was to be held in place using double-sided adhesive tape. The same methods for fixing components were used in the 580mAh version of the enclosure.

After reviewing both designs with Dr. Bonato, minor changes to the designs were made, and both concepts were built using the earlier mentioned Viper SLA machine in the lab. The pictures below show the original SHIMMER design along with the two prototypes.
It is clear, that the 580mAh battery has a significant impact on the overall size of the new enclosure.

Figure 94: Size comparison between the original SHIMMER and the two concepts.

Figure 95: Rapidly prototyped enclosures for the 500mAh and 580mAh versions of SHIMMER enclosure redesigns.
The main reasons for building the prototypes of both enclosures were to be able to physically compare the relative sizes of the two different designs, in comparison with the original SHIMMER design, and to verify the validity of certain design features. The middle and top enclosure pieces were designed to have a 0.01” gap around the perimeter, in order to create a preload on the motherboard support features. This in turn, would hold the motherboard firmly in place. With the enclosure prototypes built, they were assembled to verify this idea. The four screws were tightened to pull enclosure pieces together, and as they flexed, the support posts pressed against the motherboard from opposite sides. After some testing, it was clear that this idea was working very well. Also, the middle enclosure piece had a provision made for the slide switch’s two “wings” to fit tightly into two slots. In the 500mAh design, there was a 0.003” clearance built into the feature on both sides of the tab. The idea is shown in the picture below.

Figure 96: Clearance around the slide switch.
This configuration did not work very well, as the switch fit rather loosely into its designated spot, and was free to shift around. In the 580mAh version, the 0.003” clearance was eliminated, and this resulted in the switch fitting snugly into the slots in the tabs, eliminating any play.

At this point in the design process, it became necessary to pinpoint the exact manufacturing process, which would be used to make the new enclosures for the SHIMMER electronics. Dr. Bonato, as mentioned earlier, had a colleague at an Intel location in Ireland, who was interested in using the results of our design work to manufacture a number of units for their own purposes. In exchange, they would supply us with 60 pieces in return. Unfortunately, it was announced to us, that the rapid prototyping machine at the Ireland location was out of order. In light of the new situation, it was decided to drop the 580mAh variant of the enclosure, and continue the development of the 500mAh enclosure, implementing the lessons learned from the first prototype. At the same time, it became necessary to identify the most economical supplier, who could produce the 60 pieces out of a material with satisfactory mechanical and biocompatibility characteristics. After doing some research into potential technologies and suppliers, a number of quotes were received from a variety of businesses. The table below lists these companies along with information on the technology and material used.
Table 4: Comparison of production quotes from various suppliers.

<table>
<thead>
<tr>
<th>Supplier</th>
<th>Location</th>
<th>Material</th>
<th>Impact Resistance [J/cm]</th>
<th>Process</th>
<th>Cost per part</th>
<th>Volume</th>
</tr>
</thead>
<tbody>
<tr>
<td>Moeller D&amp;D Inc.</td>
<td>Seattle, WA</td>
<td>Urethane Resin 700</td>
<td>0.53</td>
<td>Original SHIMMER</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Robotics &amp; Mechatr. Lab</td>
<td>Boston, MA</td>
<td>Accura SI 40</td>
<td>22.27</td>
<td>Stereolithography</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Vaupell</td>
<td>Hudson, NH</td>
<td>Somos DMX</td>
<td>61.71</td>
<td>Stereolithography</td>
<td>$99</td>
<td>1</td>
</tr>
<tr>
<td>M2 Systems</td>
<td>New Britain, CT</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>$450</td>
<td>1</td>
</tr>
<tr>
<td>InterPRO Rapid Tech</td>
<td>Deep River, CT</td>
<td>Somos 9420 EP</td>
<td>44.48</td>
<td>Stereolithography</td>
<td>$60</td>
<td>60</td>
</tr>
<tr>
<td>Star Prototype China Ltd</td>
<td>China</td>
<td>PU8150</td>
<td>0.12</td>
<td>Vacuum Casting</td>
<td>$25</td>
<td>60</td>
</tr>
<tr>
<td>Accelerated Techn.</td>
<td>Austin, TX</td>
<td>Somos 18420</td>
<td>0.921</td>
<td>Stereolithography</td>
<td>$25</td>
<td>60</td>
</tr>
<tr>
<td>Moeller D&amp;D Inc.</td>
<td>Seattle, WA</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>$30</td>
<td>60</td>
</tr>
<tr>
<td>3D Proto</td>
<td>Round Lake, Ill.</td>
<td>Somos 18420</td>
<td>0.921</td>
<td>Stereolithography</td>
<td>$33</td>
<td>60</td>
</tr>
<tr>
<td>3D Proto</td>
<td>Round Lake, Ill.</td>
<td>SR-200</td>
<td>-</td>
<td>Multi-Jet Modeling</td>
<td>$20</td>
<td>60</td>
</tr>
</tbody>
</table>

The above price quotes were based on a single component of the three-piece assembly from the first prototype of the 500mAh enclosure pictured below. Since all three components of the enclosure were similar in size and shape, it was anticipated that the total cost of one enclosure would be very close to three times the figure given in table 1.

Figure 97: Screenshot of middle piece from the 500mAh enclosure, used in the quoting process.

Initially, the vacuum casting manufacturing process seemed to offer the lowest price, while the material fulfilled biocompatibility requirements, but the process’ ability to produce thin tall features was questionable. Finally, 3D Proto Inc., a rapid prototyping
company in Round Lake, Illinois, offered a very competitive production quote, using multi-jet modeling technology and SR-200 acrylic polymer. The multi-jet modeling process is a material – additive, rapid-prototyping process akin to printing. The machine’s multiple 96 jets deposit a molten photosensitive polymer, layer by layer, to form the geometry of the part. According to the company, the main uses for parts created using multi-jet modeling are aesthetic visual representations, models, small and precise prototypes, electronic enclosures, toys and figurines and prototypes for light mechanical testing.

Now that the most likely parts supplier has been identified, along with the manufacturing process and the material, the design of the enclosure was re-visited. Since the larger 580mAh variant of the enclosure was abandoned, we focused on the 500mAh design. The price of the whole enclosure is largely driven by the number of pieces in the assembly, and the possibility of arranging and constraining from motion all the electronic components using a two-piece enclosure was tried again. It was found to be possible. The pictures below show the redesigned 500mAh enclosure.
Figure 98: 500mAh enclosure – revision 1- isometric view from the top – gyro board variant.
Figure 99: 500mAh enclosure – revision 1 – isometric view from the bottom.

Figure 100: 500mAh enclosure – revision 1 – EMG board variant.
This revision featured a number of changes from the first version. As mentioned earlier, this enclosure was now comprised of two parts, instead of three. This change made it necessary to move the slide switch location to the end opposite charger interface. “ON” and “OFF” engravings were added to help identify switch position. While the original SHIMMER package was affixed to the body using an mp3 player type strap with a pouch, it was decided that this SHIMMER enclosure would be fixed to the body (arm, wrist, thigh, ankle, etc.) with a stretchy hook and loop band, without a pouch. The floor of the bottom enclosure piece was made thicker, to make room for a transverse slot large enough for a hook and loop strap to pass through. There were also lightening pockets added to the bottom of the floor to reduce material usage and part weight. To help install this strap in the bottom of the enclosure, another slot was designed in to give access and push it through, if necessary.

With the disappearance of the middle enclosure piece, there was nothing holding the battery in place. A column feature extending from the top of the cover has been added, to accept silicone compressive foam material, which would press against the battery from the top, and keep it from shifting. The same idea was used to hold the slide switch in place. An FDA compliant adhesive silicone foam rubber from McMaster was chosen due to its good mechanical characteristics and high temperature resistance. The enclosure was designed to compress the foam about 25% and apply a pressure to the battery and the slide switch. See picture below.
Figure 101: Side view of the enclosure with polyurethane foam shown in uncompressed state.

With all the design updates implemented in the CAD model, it was time to build another prototype to verify that all the parts fit properly, and that the polyurethane foam can in fact hold the battery and the switch in place. As it was done earlier, the second prototype was build using the Viper SLA system in the Mechatronics and Robotics Lab at Northeastern University. Small pieces of the compressive foam were cut with scissors, and glued to the correct areas located on the inside of both enclosure covers. The picture below shows all the parts, which together make the two enclosure variants.
Both enclosure configurations were assembled with all their respective components located inside. As it was mentioned earlier, one of the main motivations behind this project was to design an enclosure, which would eliminate any shifting of components when vigorously shaken. The greatest shock this wearable sensor package would experience would be if a subject wore it at one of their limb’s extremes, a wrist or an ankle, and performed quick “jerky” movements. To simulate this condition, a healthy, athletic young male with average upper limb-length, tested both variants of the enclosure, by holding them in the palm of his hand, and shaking them vigorously and abruptly. No shifting was detected, either via the anticipated “knocking” sound, nor were there any vibrations palpable, such as if a shock resulted in the battery dislodging from its location, crashing into a sidewall and coming to an abrupt stop. The diagram below shows how the two variants of the SHIMMER are assembled.
Figure 103: Assembly of the two variants of the SHIMMER enclosure.
The picture below compares the original SHIMMER enclosures and the new ones.

Figure 104: Original SHIMMER enclosures are shown in the top row, with a 250mAh battery. The redesigned enclosures are at the bottom, with a 500mAh battery.

With the new 500mAh lithium-ion battery, both the size and weight of the wearable sensor package increased. The table below summarizes the size and weight figures of the enclosures.
Table 5: Comparison of weights and sizes of the affected components and complete assemblies: original versus new. * These figures were calculated based on volume of the parts and specific gravity of the material.

<table>
<thead>
<tr>
<th>Affected Components</th>
<th>Weight [g]</th>
<th>Size (L x W x H) [in]</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Original</td>
<td>Redesign</td>
</tr>
<tr>
<td>Bottom</td>
<td>8</td>
<td>11*</td>
</tr>
<tr>
<td>Gyro Top</td>
<td>6</td>
<td>7.5*</td>
</tr>
<tr>
<td>EMG Top</td>
<td>6</td>
<td>7.5*</td>
</tr>
<tr>
<td>Battery</td>
<td>6</td>
<td>16</td>
</tr>
<tr>
<td>Screw(s)+nut(s)</td>
<td>0.5</td>
<td>2</td>
</tr>
<tr>
<td><strong>Complete Assembly</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Gyro-equipped</td>
<td>26.5</td>
<td>42.5</td>
</tr>
<tr>
<td>EMG-equipped</td>
<td>30.5</td>
<td>46.5</td>
</tr>
</tbody>
</table>

As can be seen in the table above, the redesigned enclosure with the new, larger battery, and additional three machine screws and nuts, collectively added extra 16 grams to the mass of the original design, with the new battery alone adding 10 grams to the overall weight. The size of the enclosure also increased. Since this wearable sensor package was meant to be worn on the wrist and the ankle, among other points of the body, it was especially necessary to keep the width and the height of the enclosure to a minimum, while a significant increase in the length was allowable. Due to the significantly larger size of the new 500mAh battery, compared to the old 250mAh battery, the width of the enclosure increased by 0.26”. By adding the transverse slot feature in the enclosure bottom piece for the strap, the height of the enclosure increased by 0.15” in case of the gyro-equipped version and by 0.21” in case of the EMG-equipped version. The lengths of both versions of the enclosure increased by 0.17”.
A.4 Manufactured Product

Now that the design of the new SHIMMER enclosure was finalized, we requested to have part samples built by the chosen supplier, 3D Proto of Round Lake, Illinois, in order to confirm the capabilities of the MJM process and those of the company. We requested that one of the parts be clear coated with an acrylic lacquer, to get a sense of the type of finish possible, and to compare the adhesive qualities between the painted and unpainted qualities. The ability of the material to accept adhesives was necessary for fixing compressible silicone rubber in place. A sample, which had been received from another company, built using the same technology, was found to be covered with waxy residue, from the manufacturing process. This made it very difficult to apply the adhesive silicone foam to its surface.

Upon receiving the samples from 3D Proto, it was found that the surface finish of the parts was very good, with no waxy residue. The silicone foam adhered satisfactorily to both the painted and unpainted surfaces. The pictures below show the samples.
Figure 105: Samples received from 3D Proto, Inc.

Figure 106: Two enclosure samples, with gyro top-cover on the left and EMG top-cover on the right.
The two variants of the enclosure were assembled, with the electronics installed. Since the parts performed as expected, an order was placed with 3D Proto Inc. for enclosure bottom (60pcs.), gyro top cover (50pcs.) and EMG top cover (20pcs.) at a cost of $2,910.00.

In order to adapt the existing strip chargers to the new SHIMMER enclosures, the top cover of the strip charger was redesigned. The images below show CAD models of the strip charger with top cover redesigned and two units plugged in.
Figure 108: Redesigned strip charger.

Rib feature ensures proper unit orientation when pluggin in for charging.

Figure 109: Redesigned charger top cover.
A.5 Discussion

Biomedical devices and wearable sensors in particular, are a product of a marriage of two engineering disciplines: mechanical and electrical, therefore their success depends on both design aspects. While today’s miniature solid-state electronic components replace the bulky, mechanical components of the past, the necessity for proper mechanical design still exists. The function of these units is to measure physical quantities, which means that the mechanical design cannot be neglected.

The redesign of the SHIMMER’s enclosure consisted mostly of adding features to better its functionality, but the main motivation behind this project was to improve SHIMMER’s mechanical interface between the subject and SHIMMER’s sensors, in order to improve collected data quality. The impact this redesign will have on the subsequent studies, could be enormous. Access to accurate, low-noise data could take place of that, which may have been deemed useless in the past. For instance, the response of the patient to different medication could now be described in a much more quantitative manner, with less time spent on “cleaning up” the data, and with trends in the data being much clearer.

A.6 Conclusion

At the onset of this project, there were four key requirements identified, which were to be incorporated into the redesign of SHIMMER’s enclosure. They were: housing a new larger battery, incorporating an on/off switch to save power, firmly locating the internal electronic components within the enclosure to avoid noise in data, and making a provision for a hook & loop type strap to affix the sensor package to the patient’s body.
There were also two limitations imposed: to keep the overall size to a minimum, and to employ the most economical manufacturing process available.

The four requirements were fully satisfied. The ability to house a 500mAh battery doubled the operating time, which should greatly improve the device’s usability. This improvement should help conduct studies like the upcoming Parkinson’s disease study, to be executed by the Spaulding Hospital.

The addition of an on-off switch will allow for greater control over the state of battery power in the units. Without a reliable way to turn the SHIMMER on or off, the battery could only be either in the process of being drained or charged, and only when the battery voltage dropped to a certain low value, the unit’s power management circuit would shut the power off. Now with a switch, units can be fully recharged and switched off, to be ready for use at any time.

One of the core functions of the SHIMMER is its ability to sense movement, using built-in accelerometers (translation) and gyros (rotation). Not only can the accelerometers detect whether movement occurs or not, but they can also assess the rate of acceleration and deceleration, or in other words the vigorousness of movement/tremor. This capability is to be utilized in the mentioned Parkinson’s disease study, to quantitatively and precisely evaluate the effectiveness of certain medications. SHIMMER’s original enclosure design was found to cause significant noise in the data, due to components shifting around. This rendered the collected data useless. The new enclosure firmly holds all components in place, and allows for full utilization of the capabilities of the electronics, greatly enhancing usefulness of the device.
Finally, the addition of a slot feature for hook and loop type strap should prove to be a more inexpensive and hygienic solution to the problem of affixing the device to a patient’s limb. The original SHIMMER enclosure was meant to fit in to a commercially available pouch or an armband, used for portable music players. This worked well, but was rather expensive. The new solution uses inexpensive hook and loop material, which can be simply replaced for hygienic reasons, if need be.

All of the features mentioned above were implemented in an envelope that is larger than the original enclosure, but the increase in size was deemed inconsequential to its functionality by Dr. Bonato. It had been expected that the proper manufacturing method for this enclosure would be some sort of an additive technology. Multi-Jet-Modeling came out on top with the best combination of quality and price.
Bibliography


