User Control of Lower Limb Prostheses

Sigurður Gísli Karlsson

Faculty of Industrial,- Mechanical
Engineering and Computer Science
University of Iceland

User Control of Lower Limb Prostheses

Sigurður Gísli Karlsson

30 ECTS thesis submitted in partial fulfilment of a
Magister Scientiarum degree in Mechanical Engineering

Advisors
Fjóla Jónsdóttir, Ph.D.
Ragnar Sverrisson, M.Sc.

Faculty Representative
Rúnar Unnþórsson, Ph.D.

Faculty of Industrial, - Mechanical Engineering and Computer Science
School of Engineering and Natural Sciences
University of Iceland
Reykjavik, January 2010

User Control of Lower Limb Prostheses
30 ECTS thesis submitted in partial fulfilment of a *Magister Scientiarum* degree in Mechanical Engineering

Copyright © 2010 Sigurdur Gisli Karlsson
All rights reserved

Faculty of Industrial,- Mechanical Engineering and Computer Science
School of Engineering and Natural Sciences
University of Iceland
VRII, Hjardarhagi 2-6
107 Reykjavik
Iceland

Telephone: 525 4000

Bibliographic information:
Sigurdur Gisli Karlsson, 2010, User Control of Lower limb Prostheses, Master's thesis, Faculty of Industrial,- Mechanical Engineering and Computer Science, University of Iceland, pp. 43.

Printing: Leturprent
Reykjavik, Iceland, January 2010

Abstract

To move and positions one´s limbs as desired is generally taken for granted, but this is not the case for lower limb amputees who can only impose their will upon a prosthetic limb by moving it. This project is concerned with significantly improving the possibilities for amputees to control a prosthetic limb as if it were a natural limb. Specifically the first goal of this project was to investigate available methods for detecting user intent and utilise them as a control signal for a lower limb prosthesis. The second goal was to select one method and demonstrate the feasibility of using it to control a lower limb prosthesis.

Four suitable methods for detecting user intent were identified, namely electromyography, mechanomyography, pressure sensing and flexion sensing. Mechanomyography(detecting sound waves caused by muscle vibration) was selected for further development. Silicone embedded sensors were constructed and mounted in a prototype socket for normal subject testing and used to control a prosthetic ankle.

The results clearly demonstrate the feasibility of using mechanomyography to control a prosthetic ankle, with an 83% movement classification accuracy To improve this system filtering techniques must be optimized. It is therefore concluded that mechanomyography is a suitable and promising technology to capture an amputee's conscious will and by extension improve his quality of life.

Útdráttur

Almennt er litið á hæfileika til þess að hreyfa og staðsetja útlimi sína sem sjálfsagðan hlut. Því er ekki að heilsa fyrir þá sem hafa misst fótlegg, eða hluta hans, en þeir geta einungis komið ásetningi sínum til gervifótar með því að hreyfa hann. Þetta verkefni snýst um að stórbæta möguleika stoðtækjanotenda til þess að stýra gervifæti rétt eins og hann væri þeirra eigin útlimur. Nánar tiltekið er verkefnið tvíþætt; fyrri hlutinn snýst um að rannsaka mögulegar aðferðir til þess að greina vilja stoðtækjanotenda og nota þær upplýsingar til þess að stýra gervifæti. Í seinni hlutanum var ein aðferð valin og hún þróuð nánar til að sýna fram á notkunarmöguleika hennar sem stýringar fyrir gervifót.

Fjórar fýsilegar aðferðir til að greina vilja notanda voru prófaðar: vöðvarafrit, greining á vöðvatitringi, þrýstiskynjun og notkun sveigjunema en greining á vöðvatitringi var valin til frekari þróunar. Hljóðnemi steyptur í sílikon var hannaður og notaður í sérhannaða hulsu til prófunar á heilbrigðum fæti. Merki hljóðnemans voru síðan notuð til að stýra gervifæti.

Niðurstöðurnar sýna svo ekki verður um villst, að greiningu á vöðvatitringi má nota til stjórnunar á gervifæti við stýrðar aðstæður. Þannig fékkst 83% nákvæmni, þ.e.a.s. gervifóturinn færðist í samræmi við vilja notandans í 83% tilfella. Hins vegar þarf að bæta merkjasíun til þess að tæknin henti við allar aðstæður. Því er ályktað að vöðvatitringur sé hentugur til þess að greina vilja stoðtækjanotanda og þar með bæta lífsgæði þeirra.

Table of Contents

List of Figures

List of Tables

Abbreviations

CNS: Central nervous system

DC: Direct current

DOF: Degrees of freedom

ECG: Electrocardiogram

EEG: Electroencophalography

EIT: Electrical impedance tomography

EMG: Electromyography

MMG: Mechanomyography (acoustic myography/AMG)

PCB: Printed circuit board

PNS: Peripheral nervous system

SNR: Signal-to-noise ratio

TF: Trans-femoral

Acknowledgements

First of all, I would like thank Össur hf. for the support provided for this project. The opportunity to work at Össur enabled me to do far more work in this project than I thought possible. The enthusiam and professional experience of numerous staff at Össur was a source of inspiration. Particularly I have to thank Andy for his encouragement throughout the project. My discussions with María Þorsteinsdóttir, Associate Professor at Faculty of Medicine, University of Iceland, provided a very useful insight into physiology and human motor control relevant for this project. I would like express gratitude to my supervisors, Fjóla Jónsdóttir and Ragnar Sverrisson for their guidance during this project. Last, but not least, I thank my one true love, Hrönn Ágústsdóttir, for her invaluable support during this project and the years of study preceding it.

1 Introduction

The most advanced lower limb prosthetic products available provide their users with unparalleled mobility and freedom when compared to older designs (Herr and Wilkenfeld 2003) but compared to able-bodied persons with fully functional lower limbs, amputees still have severe mobility limitations in their daily lives. The work presented here aims to take a significant step towards reducing those limitations. The number of amputees living in the USA alone was estimated at 1.6 million people in 2005 and it is expected to rise significantly in the coming years (Ziegler-Graham, MacKenzie et al. 2008), so this is a problem of great proportions.

For a prosthetic device to function as a natural limb does, it is necessary for a control system to match the motor control system of the human body. Despite advances in the field of artificial intelligence, independent prosthetic control systems are still far from reaching that level. Therefore it is suggested that connecting directly to the existing human motor control system of an amputee can significantly improve the state of the art in lower limb prosthetics.

Much research in this area is focused on upper limb applications but this project will focus on lower limb applications exclusively, as this is a neglected area of research with great potential for technological advancement. Specifically, the following research questions will be addressed:

- What type of available sensors can be used for obtaining voluntary control of lower limb prostheses?
- Can the selected technology provide sufficiently accurate and reliable information for lower limb prosthetic control?

In order to find a connection between the human motor control system and a prosthetic device, an understanding of both systems is necessary and the following chapter of this thesis therefore provides a concise overview of locomotive control and prosthetic control systems. This is followed by the introduction of three different methods of merging these systems. A literature review of muscular activity detection for control purposes follows, as detecting muscle activity is deemed the most feasible method of detecting user intent for this project. Several different sensor technologies for this purpose are tested and evaluated in chapter four. Detecing pressure fluctuations caused by muscle vibrations, known as mechanomyography (MMG), is selected and a suitable sensor is developed, based on existing literature of upper limb applications, in chapter five, which also deals with finding suitable muscles for detection and control. To test the developed sensor at the selected site, for lower limb prosthetic control, a prototype for normal subject testing was constructed. As described in chapter 6 this prototype consists of two pairs of electret condenser microphones embedded in silicone. The sensor pairs are fitted in a normal subject version of a prosthetic socket, and used to control a prototype prosthetic ankle. This facilitates the testing of MMG for lower limb prosthetic control, which has not previously been described in the literature. The test results, shown in chapter seven, indicate that MMG can be used for control of lower limb prosthetics in the swing phase of gait, but a further development of the embedded sensor and filtering techniques is needed for succesful control in all types of gait activitity. Suggested methods for achieving this are outlined in the final chapter.

2 Locomotive and Prosthesis Control

2.1 Human Locomotion Control

Human intentional control of the lower limbs originates in the central nervous system (CNS), although many aspects of this control system are unknown and controversy exists regarding the respective roles of the brain and spinal cord (Yang and Gorassini 2006). Multiple theories of motor control have been suggested (Shumway-Cook and Wollacott 2007) but no single theory can completely explain all elements of human locomotion. For the purposes of this project it can be assumed that locomotion control resides in the CNS and that the systems inputs include vision, balance and proprioception. A proposed schematic of this system is shown in *Figure 1*. Although feedback is an important aspect of locomotion control, central pattern generators in the CNS (i.e. pre-programmed movements) and feedforward control (responses to anticipated movements) also play an important role. The entire system is also highly adaptive to new situations, a phenomena described as neural plasticity. In normal gait, the response time of the system is of a lesser importance, as the muscle activation is repetitive and predictable. In the case of external perturbations, e.g. stumbling over a threshold, a very rapid reaction may be needed to prevent falling. Nashner (1977) reported a 100-120 millisecond delay from perturbation to muscle reaction. The reaction time of a muscle sensor-actuator system will unavoidably lenghten this reaction time.

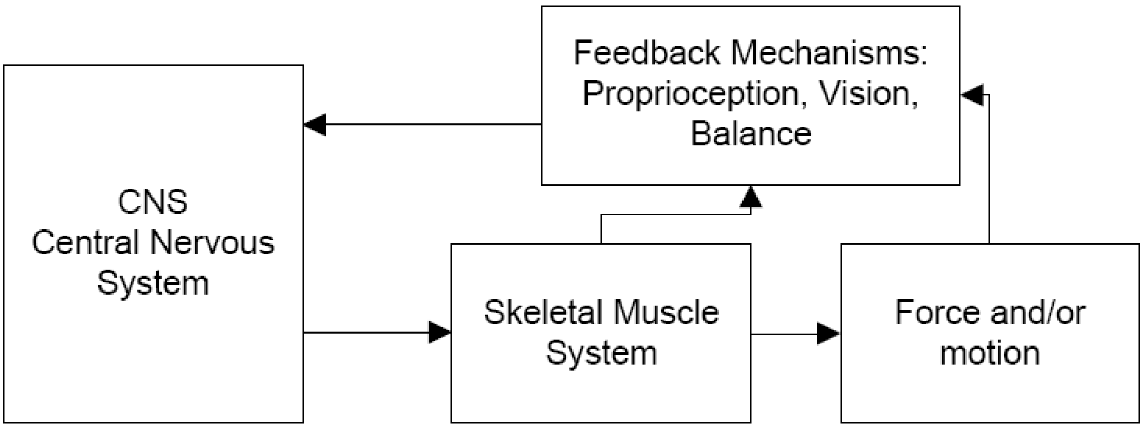


Figure 1. A simplified view of human locomotion control.

2.2 Prosthetic Locomotion Control

Current advanced lower limb prosthetics are controlled by microprocessors, using pre-programmed algorithms or artificial intelligence to predict an appropriate response to environmental situations. The environment is sensed by different sensor technologies; accelerometers, load cells, anglometers, gyroscopic sensors etc. These currently comercially available systems operate independently of the user‘s intent, although the user can affect the system by moving or loading the prosthesis, as shown in *Figure 2*.

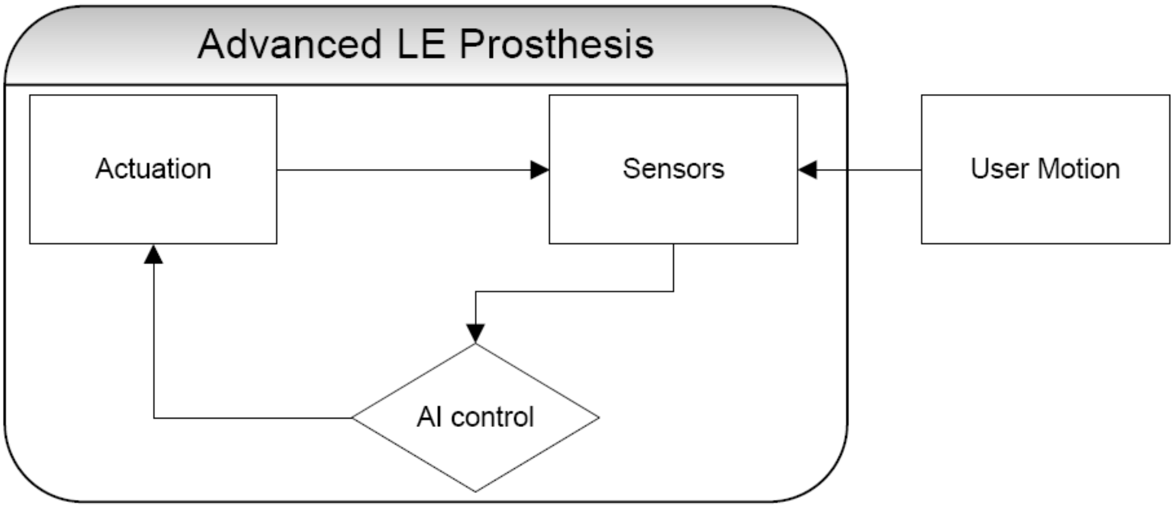


Figure 2. Control architecture of commercially available advanced prostheses. The user can affect the system indirectly, e.g. by loading the prosthesis or moving the residual limb.

2.3 Using User Intent for Prosthesis Control

In this project, amputees, prosthetists and prosthetic engineers were interviewed to establish what type of control is required, what would improve current prosthetic products, and what types of sensors might be promising for prosthetic control. This investigation revealed that current advanced lower limb prosthetics are generally very good for a range of “standard” gaits such as walking level ground, on slopes, up and down stairs (these are well known patterns and do not provide a problem for microprocessor control). However, transition between states and non-gait activity (e.g. washing a car) can be troublesome in most cases. This means that a user intent-based control would be utilised to optimise well-defined gait patterns, but crucially it is critical in unusual or unexpected situations, such as stumbling, or performing precision control activities (e.g. kicking a football)

Several different methods of using user-intent for prosthetic control can be utilised. With current microcomputers, a prosthesis can be programmed to always perform a certain set of functions in a corresponding situation (state), and then use the user intent signal to switch between the pre-programmed states. By programming sufficient states, most or all situations encountered by amputees can be dealt with in this manner. A diagram of this system is shown in *Figure 3 a)*.

Another method would be to focus on the amputee´s remaining muscles, used for locomotion and process the signals from those with advanced pattern recognition algorithm to predict intended action. *Figure 3 b)* shows this control architecture.

A third method could be to focus entirely on neural/brain plasticity and use single muscle signals for each degree of freedom (DOF) of the prosthetic limb, i.e. a certain muscle will always activate the same function, as described in *Figure 3 c)*. This means that an amputee must learn to apply the muscles at correct times for a multitude of different situations. Although this learning period is definitively a disadvantage, the more direct control can allow the amputee to adapt to all situations. This method is focused on when selecting and developing sensor technology in this work.

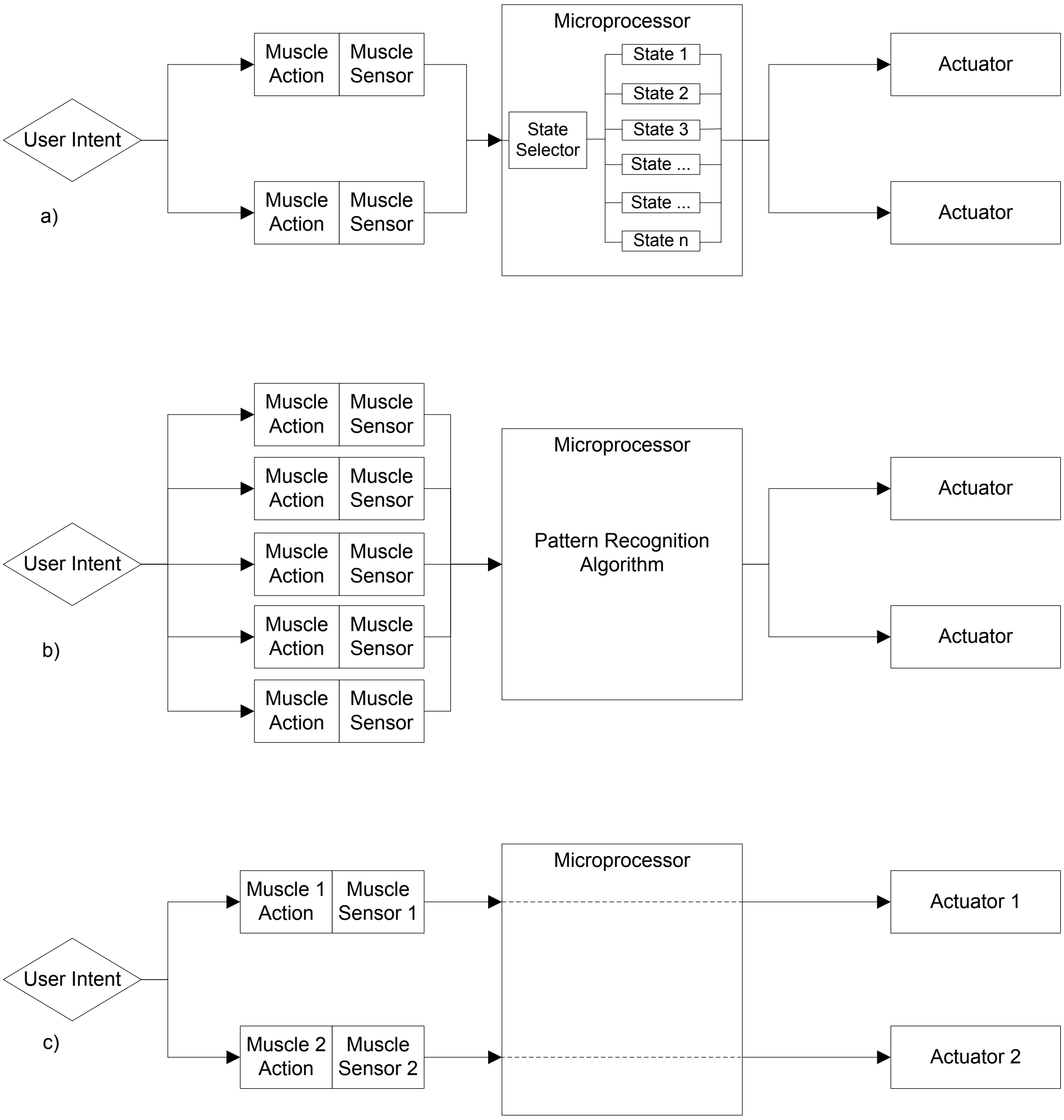


Figure 3. Proposed methods for user control of prostheses. a) Muscle signal for selecting gait state b) pattern recognition of muscle signals c) One-to-one control relationship between muscles and DOF.

3 Review of Prosthetic Control Literature

3.1 Overview

Before selecting sensor technologies for investigation in this project a literature review was carried out, to determine the state of the art and identify suitable technologies. One of the research areas is electroencophalography (EEG) or the detection and interpretation of human intent from brain waves or neural signals. EEG has been researched extensively (Lebedev and Nicolelis 2006) but a prosthetic control system based on those technologies is unlikely to become available for some time, due to limitation in the robustness and life span of the available invasive sensor technology. Due to the invasive nature of these technologies, considerable health, safety and ethical procedures are required for experimentation, these factors ensure a timescale beyond the scope of this project. This project is therefore restricted to state of the art non-invasive techniques, mainly focused on detecting muscle activity. Methods for detecting muscle activity include electromyography (EMG) and mechanomyography (MMG) but voice recognition systems have also been used for prosthetic control.

3.2 Electromyography

EMG is widely used in clinical studies and is the preferred tool for non-invasive muscle activity monitoring. EMG may be invasive (fine-wire inserted directly into the muscle) or non-invasive (signal detected from skin surface). Most studies involving EMG occur in a short-term clinical setting and as such sparse literature is available regarding long term use. Konrad (2005) provides an overview of common methods used to counter inter-subject variability, and day–to-day variability. Problems with EMG signals as a control for a prosthesis include changes in skin impedance (e.g. due to sweating), a need for location accuracy (as the muscle may move under the skin) attachment issues due to hair or external pressure (e.g. sitting on electrode).

Several researchers (Arieta, Katoh et al. 2006; Kato, Fujita et al. 2006; Kondo, Amagi et al. 2008) focus on using EMG for controlling multiple DOF arm prosthesis and an EMG controlled arm prosthesis is commercially available (Otto Bock 2009). There are however a few differences between controlling upper and lower extremities that are worth noting. First, a human hand has more than 20 DOFs, requiring a much more sophisticated control system to control a dexterous prosthetic hand than a prosthetic leg, which typically has only one or two DOF (although the human leg arguably has four main DOFs below the hip). This would indicate that controlling a simple prosthetic leg should be an easier task than controlling a prosthetic hand. However, the larger forces involved in human locomotion pose a challenge to the mechanical integrity of the surface EMG sensors and there is need for very rapid reaction as mentioned in section 2.1, to prevent falling in unexpected situations. Although some controversy exists in the literature, Farrell and Weir (2007) report an optimal controller delay for a prosthetic hand at 100-125 ms, as the best compromise between pattern recognition accuracy (requiring a longer signal) and responsiveness (requiring fast reaction to input). This optimal controller delay is presumable considerably shorter for lower limbs.

The use of EMG as a control signal for lower limb prosthetics can be classified into two different segments. Most researchers focus on using EMG signals to control finite state control schemes, detecting whether the user is walking on level ground, up/down ramps or stairs, etc. This includes recent work on an active prosthetic ankle by Au, Berniker et al. (2008) and other pilot studies (Jin, Yang et al. 2006; Huang, Kuiken et al. 2009) as well as older work by Peeraer, Aeyels et al. (1990) and Aeyels, Van Petegem et al. (1995). The latter reported very successful trials with three amputees on a magnetic particle knee brake controlled by EMG and other sensors. Few have reported direct control of a prosthesis with an EMG signal, but Myers and Moskowitz (1981) reported a successful proportional control of knee torque in a fixed leg laboratory setting with a single amputee, using seven EMG signals and Horn (1972) used EMG pulse signal from a redundant stump muscle to activate a magnetic on/off brake. No reports of long-term, multi-subject testing were found. It is suggested that this may be caused by a lack of success in long term trials, a shift in research trends towards upper limb applications or finite state models, or simply because actuator and sensor technology available at the time was too limiting. It is proposed that recent advances in AI control of lower limb prosthetics, and the widespread use of EMG in clinical settings and upper limb prosthetics, may have changed this, warranting further research of proportional or on-off EMG control of lower limb prosthetics. This is further supported by recent work on long-term usage of EMG electrodes (Garcia, Zaccone et al. 2007).

3.3 Mechanomyography

When muscles contract they produce a resonance frequency vibration, which can be recorded as sound. This is known as mechanomyography (MMG) or as acoustic myography (AMG). The main frequencies of these sounds are at 5-50Hz, with a power peak at 15-18 Hz (Grass Technologies, 2009). The RMS value of this muscle sound has been shown to be proportional to the muscle effort (Barry, Geiringer et al. 1985; Courteville, Gharbi et al. 1998) and it has been used to monitor muscle fatigue (Al-Zahrani, Gunasekaran et al. 2009). Several recent papers report the simultaneous use of MMG and EMG (Cramer, Housh et al. 2004; Coburn 2005; Ebersole, O'Connor et al. 2006), but most of the results are based on clinical settings and even stationary measurements which is not sufficient for prosthetic control.

Using MMG for controlling a prosthesis was suggested by Barry, Leonard Jr et al. (1986) where a free-standing single DOF prosthetic hand was controlled by MMG signals. Two test subjects were able to open and close the hand within a three-minute learning period. Two US patents (4571750 and 4748987) were filed but there are no reports of commercial products based on this work.

Silva, Heim et al. (2005) describe a self-contained MMG controlled prosthetic hand with a 120 millisecond delay from intent to action. The two test subjects were capable of 88% and 71% control accuracy, respectively. A detailed description of a coupled accelerometer-microphone sensor used in this study is provided by Silva (2004) as well as a mathematical model for signal processing and a classification strategy for prosthesis control. Plans for utilization at the Bloorview McMillan hospital do not seem to have been realized.

3.4 Other Technologies

Apart from microprocessor or artificial intelligence control, EMG and MMG are the most common user control strategies for prosthetics found in literature. Mainardi and Davalli (2007) suggest using a custom built throat microphone (laryngophone) and a commercial voice recognition system. Tests on two healthy subjects have revealed 97% classification accuracy. It is demonstrated that this may be utilized to reduce the time required for complex reach-and-grasp tasks with an arm prostheses but the suitability of voice commands for a lower limb prosthetic control system is questionable.

4 Selecting Sensor Technology

4.1 Sensor Brainstorming

Suitable technologies for prosthetic control need not be limited to those described in the literature review above. To find and classify other available methods a mind map was constructed during the idea generation phase of the project (*Figure 4*). Detecting physical muscle change can be achieved measuring pressure or force (between prosthetic socket and skin) or by measuring geometrical changes of the muscle, such as displacement, circumference or shape (contour) change. Muscle activity can be measured with MMG, EMG, ultrasound or even electrical impedance tomography as suggested by The Open Prosthetics Project (http://openprosthetics.org/). Other abstract ideas include detecting intent via an eye direction sensor or with a simple control panel operation, both having the major disadvantage of occupying other body parts than normally used for locomotion. A combination of more than one sensor technology can be used to eliminate characteristic drawbacks associated with each technology.

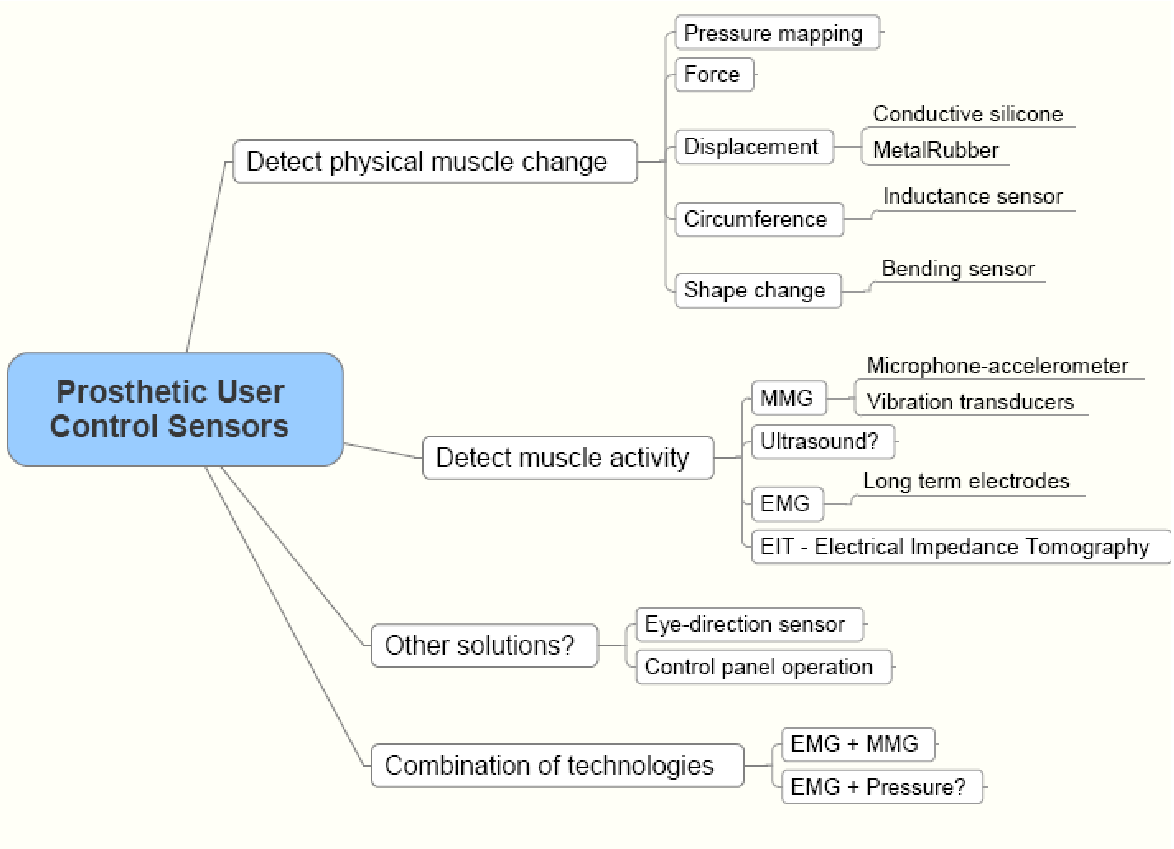


Figure 4. Mind map from user control sensor technology brainstorming.

The selection of sensors for preliminary testing was constrained to sensors that were either simple to build or commercially available (and relatively inexpensive). Measurements were carried out with EMG, MMG, a pressure sensor, an inductance sensor and a flexion sensor.

4.2 EMG Measurements

Drawbacks of EMG as a control signal for prosthetics include lack of repeatability, which may be caused by day-to-day variability, person to person variability and the need for location accuracy, as well as susceptibility to electrical noise. To quantify the scale of these issues two tests were carried out.

4.2.1 EMG Repeatability Testing

EMG measurements were carried out on two normal 30 year old male subjects. Subject 1 was fitted with short-term adhesive electrodes. These electrodes needed replacement after exercising and showering and simple skin markings were used for location accuracy. Subject 2 was fitted with long term EMG electrodes, intended for electrocardiograms (ECG), during the entire test period (three days). The electrodes were placed proximally on the *peroneus* muscle (high on the outside of the calf), see *Figure 5.*

Figure 5. Position of EMG electrodes. Long term electrodes (left) and short term electrodes (right).

The testing was carried out using wireless EMG equipment (from Kine ehf.) connected to a PC with a data collection and visual feedback program (Kinelive). The equipment is depicted in *Figure 6*.

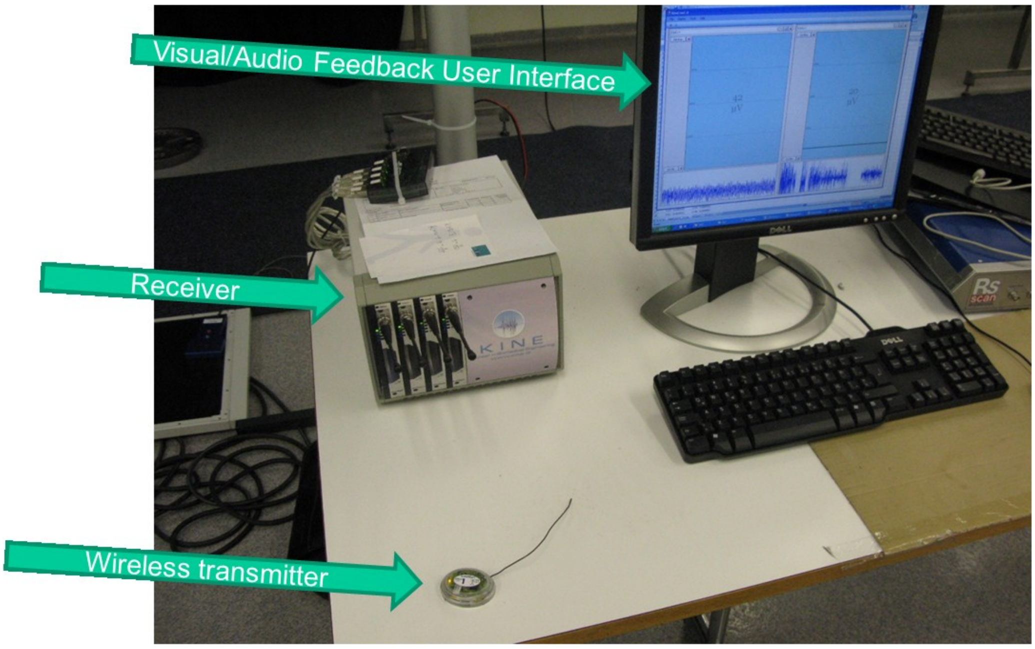


Figure 6. EMG equipment used in testing.

A special test protocol was repeated during every test. Subjects were instructed to use the visual feedback supplied to follow this protocol:

- 0-10 sec: Relaxed in sitting position.
- 10-30 sec: Maintain an EMG signal strength of 25-35 µV.
- 30-40 sec: Relaxed in a sitting position.
- 40-50 sec: Rise to a standing position and maintain it.
- 50-55 sec: Perform a maximal isokinetic contraction, guided by the visual feedback.
- 55-60 sec: Relaxed in a standing position.
- 60-80 sec: Normal walking on level ground.

The data obtained over the three days of measurements is shown in *Figure 7* and *Figure 8*. The main conclusions from these measurements are:

- A repeatable signal can be produced with some accuracy (10-30 sec.) with little or no training if feedback is available.
- There is significant day-to-day variability in maximal contraction and normal walking, especially for subject 2.
- The subject-to-subject variability is less than the day-to-day variability.

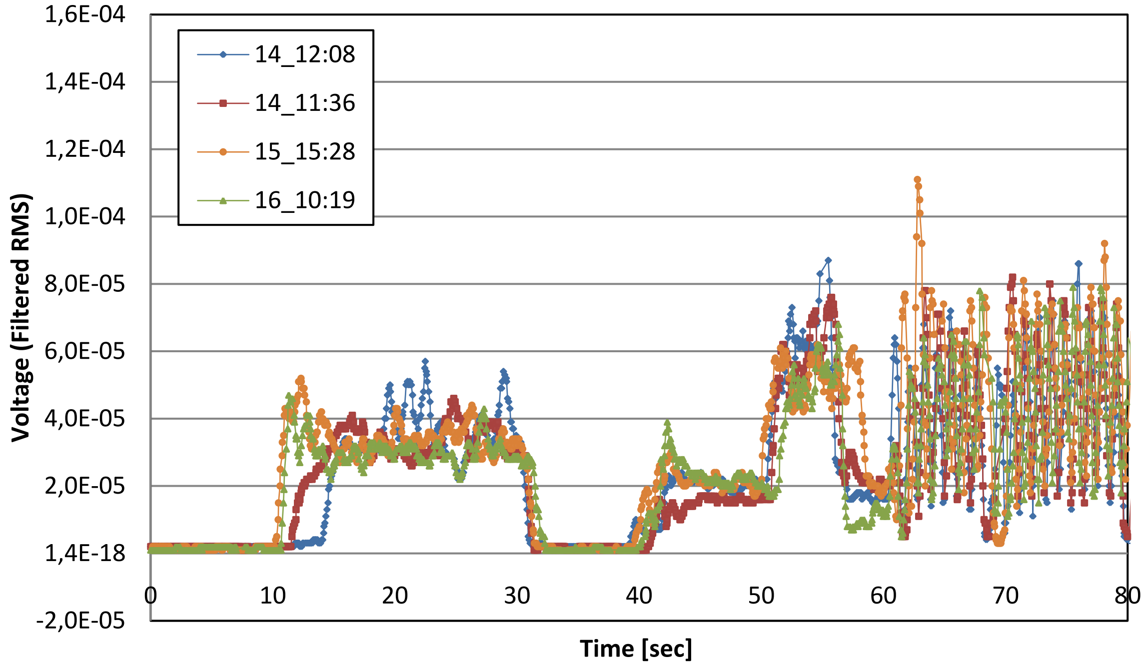


Figure 7. Repeated EMG measurements (subject 1). The legend format is date_time (dd_hh:mm).

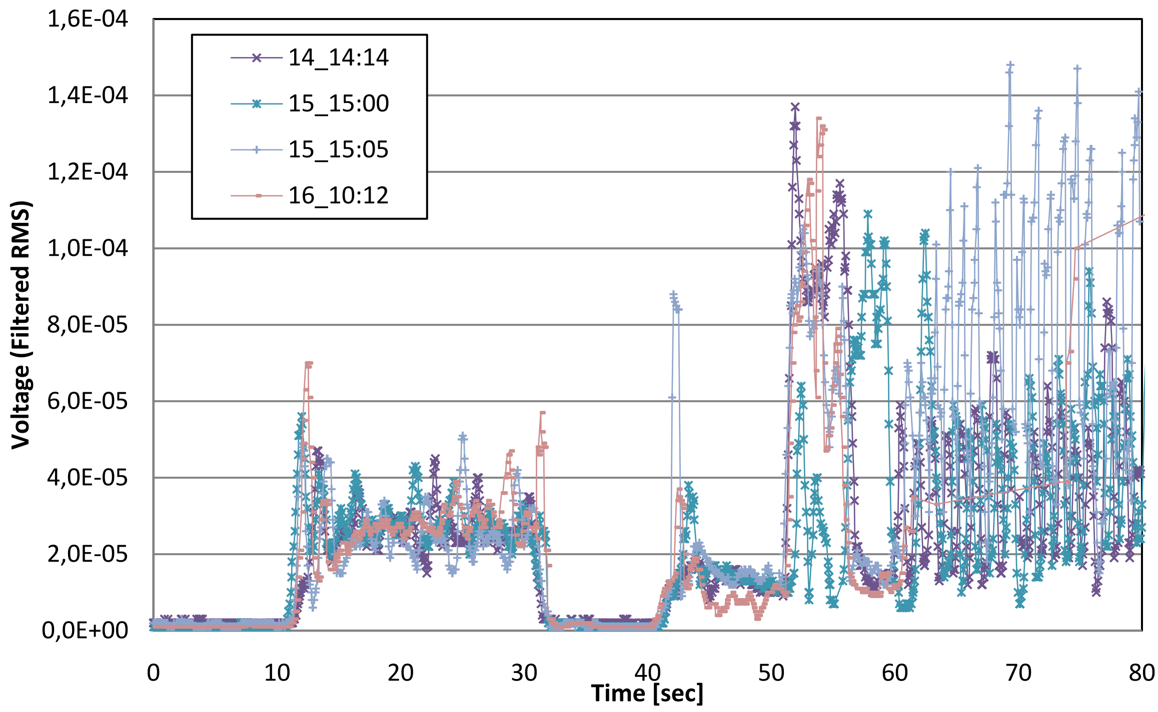


Figure 8. Repeated EMG measurements (subject 2). The legend format is date_time (dd_hh:mm).

4.2.2 EMG Noise Sensitivity

Since the EMG signal is on the µV scale, electrical noise may produce significant interference in the measurements. To investigate this, a simple noise test was carried out. The following test protocol was repeated twice, with an additional transmitter not connected to an EMG electrode in the later trial (red line):

- 0-10 sec: Relaxed in sitting position.
- 10-15 sec: Maximal isokinetic contraction.
- 15-25 sec: Relaxed in a sitting position.
- 25-40 sec: Tapping the transmitter a few times and waiting.
- 40-70 sec: Activating a 1500W, 220VAC device (kettle) within centimetres of the electrode/transmitter pair.
- 70-80 sec: Relaxed in a sitting position.

The results are shown in *Figure 9*. It can be seen that the small impacts on the transmitter have an amplitude similar to the maximal contraction, but also that the AC current has only a minor effect in one of the trials (and the interference is suspected to be caused by unintended movement). This correlates well with the high stated common mode rejection ratio of the Kine equipment.

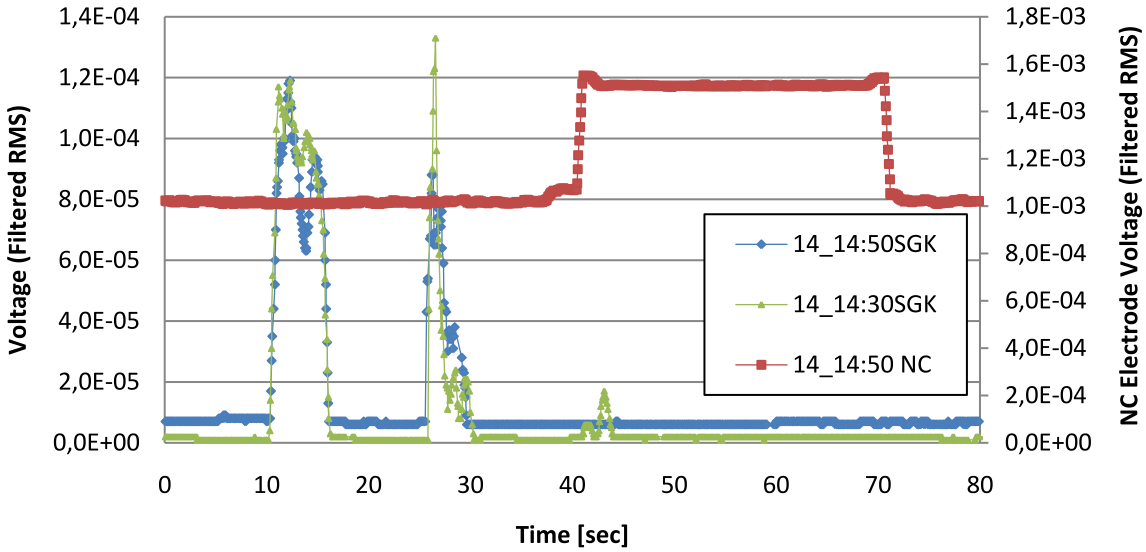
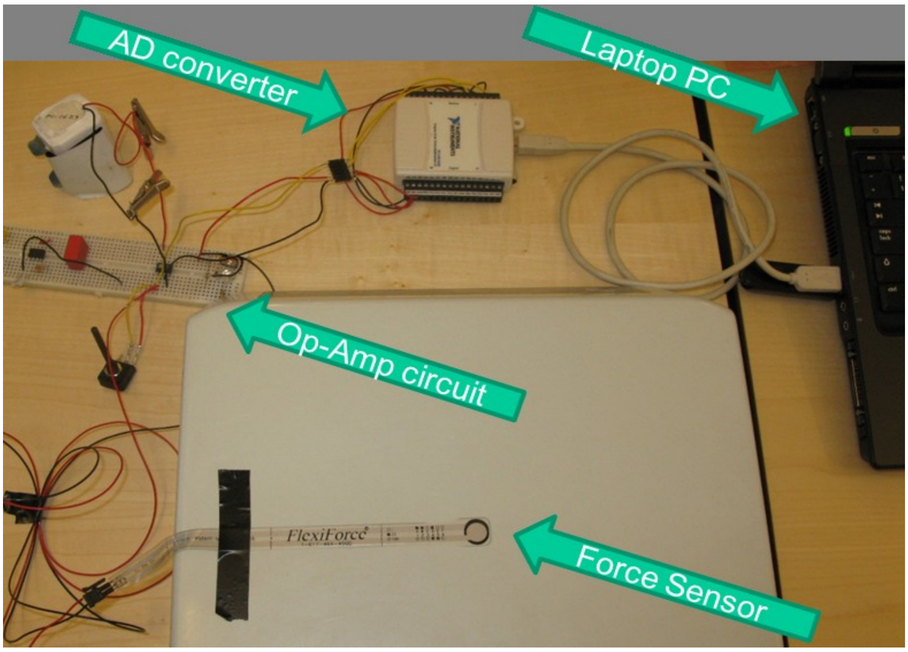


Figure 9. Noise in EMG signals. The red line is from a transmitter not connected (NC) to an electrode (right hand side axis).

4.3 Force Sensor Measurements

Measuring the force generated by a muscle and transmitted through the skin is one method of attaining a voluntary control signal from an amputee. The force can either be measured inside the prosthetic socket or in a different location on the amputated leg (below knee amputees) or the sound leg (above knee amputees). Measuring inside the prosthetic socket has the benefit of having a hard surface on one side of the force sensor, creating a stronger signal. The drawbacks of measuring inside the socket are that residual muscles differ greatly between individuals and muscle atrophy is common due to lack of use. Chemical resistance of sensor material is also an issue as amputees usually sweat inside the liner and shear stresses from in-socket motion can also damage the sensor. For these reasons a paper-thin pressure sensor called Flexiforce from Tekscan Inc. was selected for testing. The sensors are based on resistive ink technology and the pressure range and sensitivity can be adjusted with an operational amplifier circuit with a variable resistor as suggested by the supplier. *Figure 10a)* shows the sensor and the data collection equipment and *Figure 10b)* is a schematic diagram of the system. The muscle force changes the resistance of the force sensor and this change is amplified and converted to a voltage by the operational amplifier circuit. An analog-to-digital (AD) converter digitalizes the analog voltage and a PC stores the data.

a)



b)

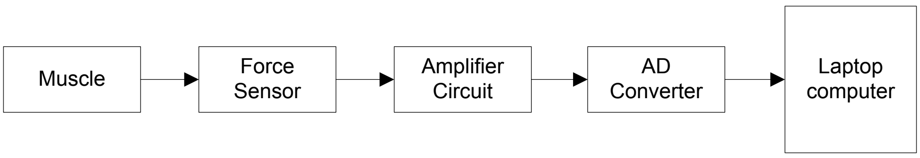


Figure 10. Force sensor and data collection equipment a), and schematic diagram of the measurement system b).

4.3.1 Force Sensor Normal Subject Testing

For testing on a normal subject, the sensor was placed between a piece of sheet metal and a cutout from a prosthetic silicon liner. The assembly was then taped to the *rectus femoris* (front of thigh), with the silicone patch facing inwards. The test protocol used is shown above the data displayed in *Figure 11*. First, there are two maximal contractions while sitting, then a single contraction in a standing position, level ground walking (at 35-60 sec), followed by a 10 second standing still period. Finally, the graph shows walking with an additional voluntary contraction in every third step.

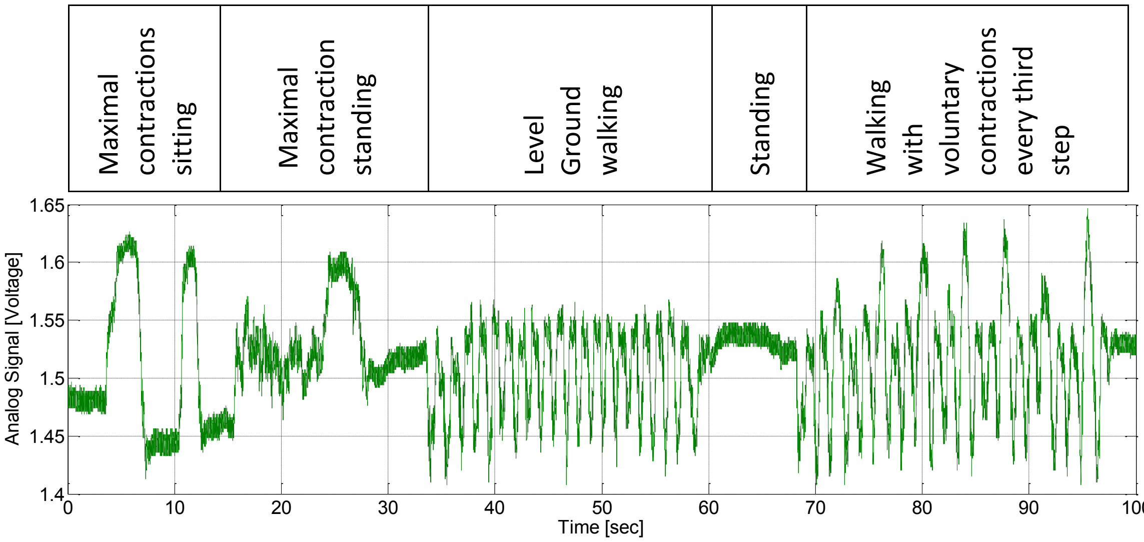


Figure 11. Normal subject testing with a force sensor.

It can be seen that voluntary contractions always have a larger signal amplitude than involuntary signals, even when considering "involuntary" activities such as walking. If this signal is used the user can both use the *rectus femoris* muscle for normal gait activities (smaller amplitude) and for a prosthetic control (larger amplitude). This indicates a feasibility of using this signal to control a prosthesis, provided that the sensor can be worn constantly as described in this setup.

4.3.2 Force Sensor Amputee Testing

Since the force sensor measures force between two surfaces it may be better suited for measurement inside a prosthetic socket, since the socket provides a rigid surface for the sensor. The force sensor was therefore also tested inside the socket of a trans-femoral (TF) amputee. The sensor was placed inside the hard socket, facing the *semitendinosus* and *biceps femoris* muscles (back of thigh). The sensitive part of the sensor was placed about 10 cm from the top of the socket, thus avoiding the need for wires inside the socket. The collected data from a single setting is shown in *Figure 12*. The test consisted of walking on level ground, then up and down a few steps, walking on level ground again, and finally standing still, with a maximal voluntary contraction at the end. It can be seen that the operational amplifier in the circuit becomes saturated during normal walking and stair walking. This is however not significant as the maximal voluntary contraction produces a much weaker signal than normal walking, thus indicating that it is difficult to use this signal from this muscle to identify user intent.

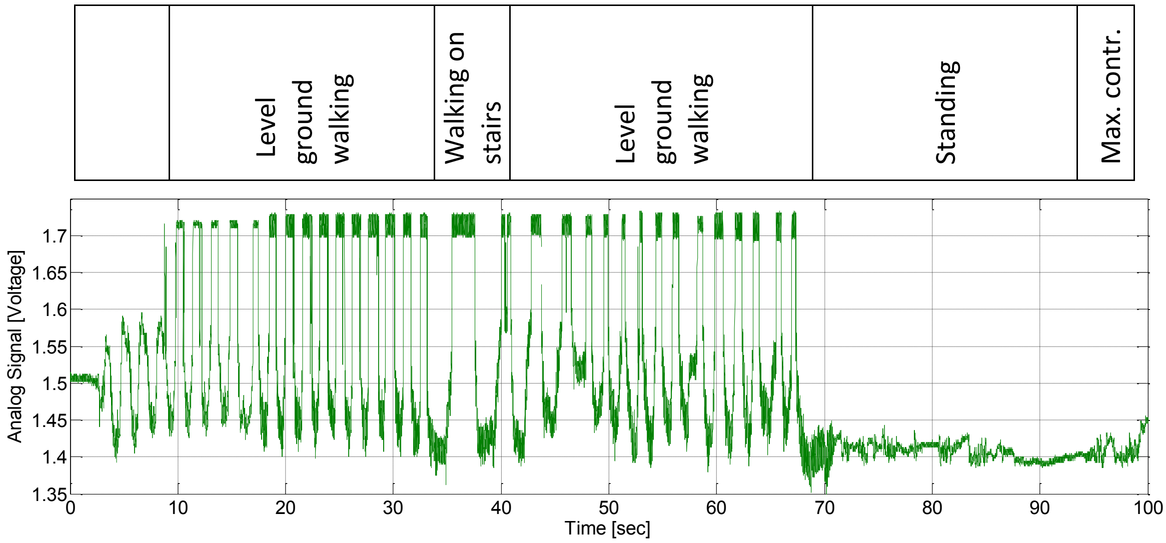


Figure 12. Force sensor on back of thigh, TF subject.

To achieve a voluntary user signal that can be distinguished from signals from normal walking the force sensor was moved to the *adductor longus* (inside of thigh). In this location a voluntary signal stronger than those created by walking could be obtained, as shown in *Figure 13*. The difference however, was less than 50 mV but this could be increased by locating a more suitable anatomical site or optimising the amplifier circuit. This pressure sensor can therefore be used to capture user intent and it has good potential as a control signal for a prosthesis.

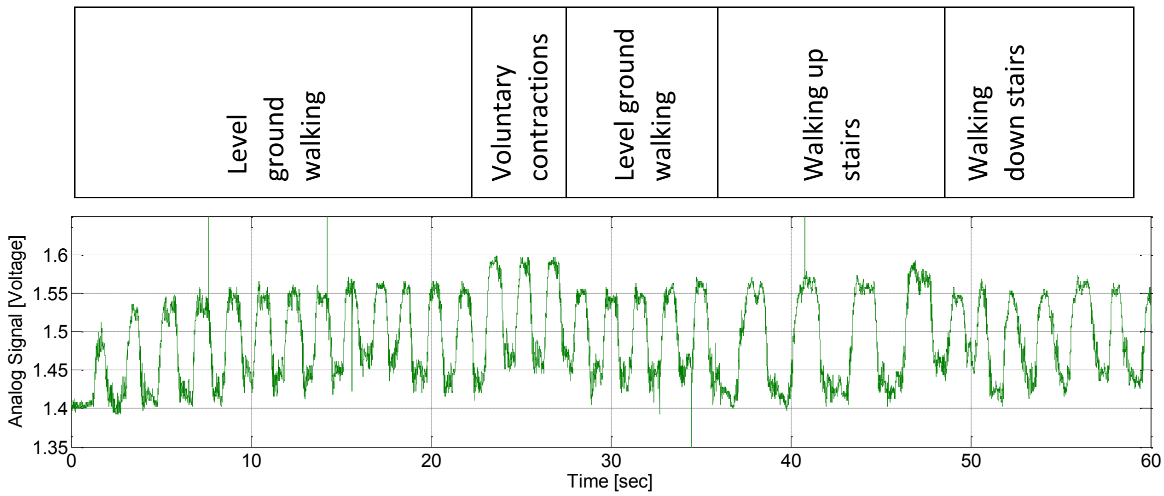


Figure 13. Force sensor on the adductor longus, inside a prosthetic socket.

4.4 Flexion Sensor Measurements

Flexion sensors utilised in this project are based on resistive ink technology (supplied by Flexpoint Inc.). A layer of resistive ink is positioned on a polyimide substrate, such that when the sensor is flexed, the ink layer is either compressed or stretched; thereby changing the resistance of the layer; see *Figure 14*. As a control signal for a prosthesis, the flexion sensors can either be used for sensing shape changes of muscles under the skin or the flexion of a joint.

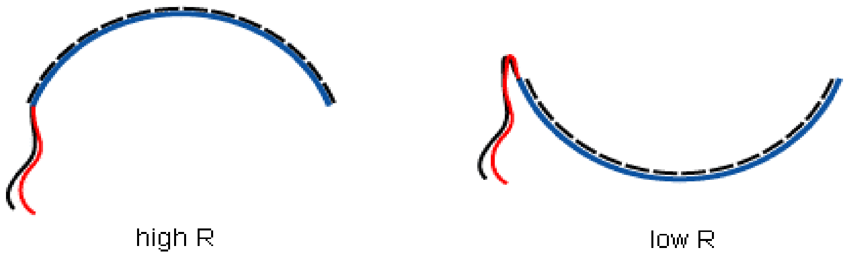


Figure 14. Working principle of resistive ink flexion sensors.

4.4.1 Muscle Shape Change

For detecting muscle shape change, the *semitendinosus* muscle was selected, as the shape change is large. Placement can be seen in *Figure 15*; the sensor is connected to an op-amp, AD-converter and a laptop just as the force sensor, described previously. This scheme creates a portable setup.

Figure 15 Placement of a flexion sensor for detecting muscle shape change.

The signal amplitude during level ground walking is about 20mV, compared to a 30mV reading from voluntary contractions, indicating that it may be difficult to get a clear control signal from this muscle with the flexion sensor.

4.4.2 Joint Flexion

It may be possible to control a prosthesis by using the movement of remaining joints. Using a finger joint for control seems a straightforward solution, but it is likely that occupying one hand for prosthetic leg control is unacceptable for most amputees. Therefore, using a toe of the intact leg was selected for testing joint movement sensing. The setup is depicted in *Figure 16*, with the op-amp, AD-converter and laptop PC.

In a stationary setting a very clear signal with amplitude of about 150 mV can be obtained. A comparison between signals from level ground and stair walking to voluntary contractions can be seen in *Figure 17*. During walking, the toe joint is not moved much, but some extension is hard to avoid during toe-off. This is however, not apparent in the observed signal as the ink layer on the sensor is much more sensitive to stretching (flexion) than compression (extension). The signal from the voluntary flexion is generally much larger than the signal from walking, but some large amplitude spikes were observed during walking. This may be caused by interference between the sandal worn during testing, and the sensor, suggesting that the sensor must be packaged properly to avoid spikes and other noise. Since these disturbances can be dealt with by either sensor packing or filtering, the flexion sensor can be used to measure toe joint bending. The sensor signal can be used for controlling a prosthesis, but successful user control depends on whether the voluntary toe bending disturbs the natural gait of the amputee or requires a high level or cognitive effort.

Figure 16. Flexion sensor setup for detecting toe joint flexion.

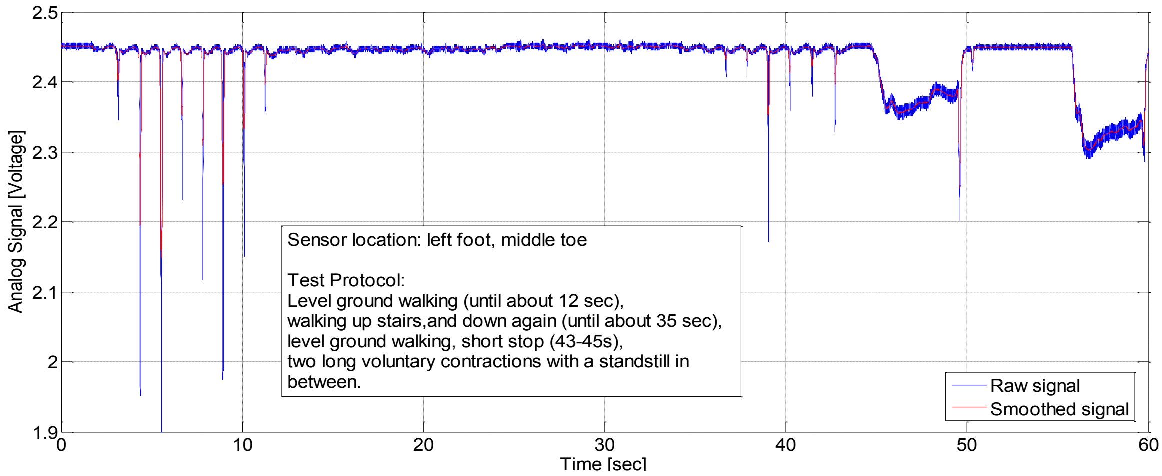


Figure 17. Flexion sensor measurements of a toe joint.

4.5 Inductance Sensor

An inductance sensor was used to measure muscle activity. The sensor is a thin copper wire braidered in cloth strip in a „sawtooth wave pattern“, shown in *Figure 18*. The cloth strip is then attached in a closed loop to the surface being measured and connected to an oscillating circuit. The coil formed by the braidered wire will affect the circuit resonance frequency, which varies as the cloth strip is stretched or compressed. This effectively means that the sensor can be used for measuring the area of the closed loop cloth strip. The output in this setup is a high frequency square wave.

Figure 18. Inductive sensor cloth strip. A thin coated copper wire is braidered in the cloth.

This sensor was attached to the thigh of a normal subject and a digital oscilloscope used for measuring the resonance frequency of the circuit. An example of the oscilloscope output is shown in *Figure 19*.

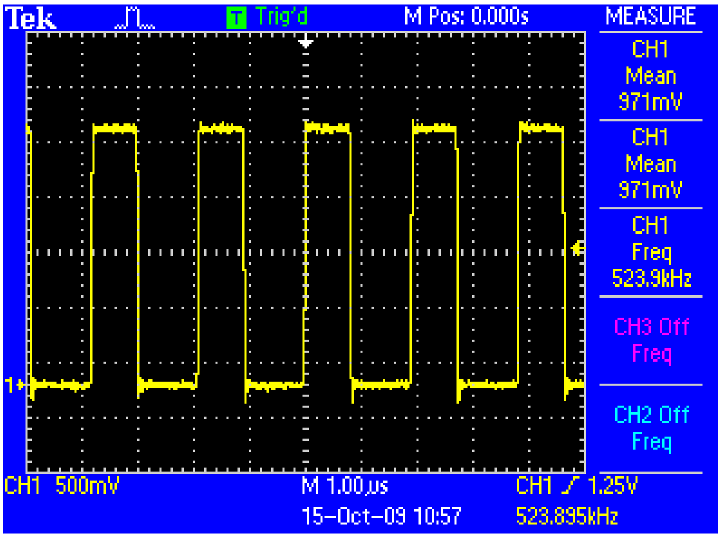


Figure 19. Oscilloscope output of an inductive sensor around a thigh in a relaxed sitting position.

The measured frequency was above 500 kHz and the noise level was about 400 Hz, which is relatively high, compared to “amplitudes“ of about 1, 2, and 3 kHz for sitting contractions, standing up, and bending knee, respectively. Since this sensor requires a rather complex drive circuitry and a small signal-to-noise ratio was observed, no further testing was carried out.

4.6 MMG Sensor

A detailed description of how to build a coupled microphone-accelerometer sensor for MMG is available online (Silva 2007). This description is based on Silva (2004) and it was followed to build a preliminary sensor for testing. The microphone used was a PVM 6027-2P423 from Veco Vansonic instead of the suggested microphone. The microphones have similar specifications. The sensor was then embedded in silicone as depicted in *Figure 20*. In front of the microphone there is a small air chamber, enclosed by a thin silicone membrane. The air chamber and membrane are used to passively amplify the signal as acoustic pressure radiated in the air by the muscle is very low (Courteville, Gharbi et al. 1998).
The silicone used for the membrane was of type LSR5850 from Nusil with durometer A50 and the remainder was conventional dental silicone (Dental ADS 931). The two parts were glued together with silicone glue from Wacker. When placed on the skin above a muscle the embedded sensor shows an amplified signal compared to a non-embedded sensor, as detailed by Silva (2004). This microphone/accelerometer couple is connected to an AD converter and the signal is captured by a Matlab program (*Figure 21*).

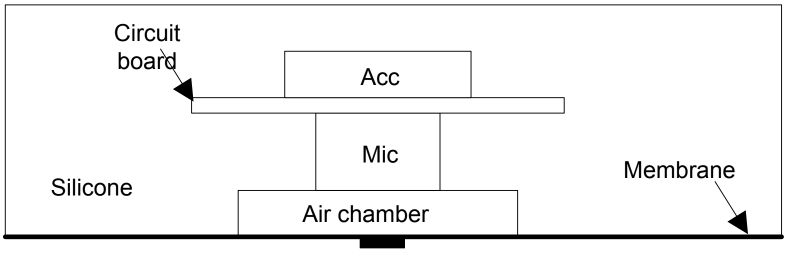


Figure 20. Microphone/accelerometer couple for MMG recording.

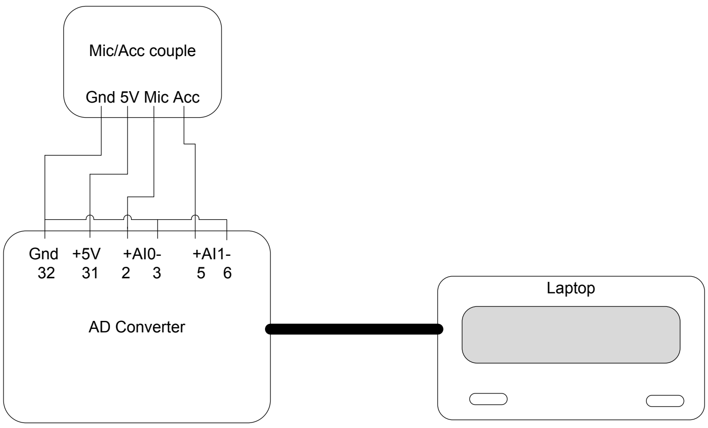


Figure 21. MMG sensor setup. The signals from the microphone/accelerometer couple are digitized by an AD-converter and read by the Matlab program on a laptop computer.

4.6.1 MMG Testing before embedding

The sensor was tested before embedding by using a short piece of plastic tubing around the microphone, creating a small air chamber in front of the microphone when it is placed against the skin. The sensor was then placed on a bicep muscle. This produces the signal shown in *Figure 22*. Rapid movements of the arm (i.e. sensor motion and muscle contraction) cause the first disturbance and the latter two are from isokinetic contractions of the bicep muscle. Both sensors capture the motion, but the voluntary contractions are only registered by the microphone. The microphone motion signal is significantly stronger than the signal from the muscle contractions, warranting the embedding of the sensor to passively amplify the muscle signals.

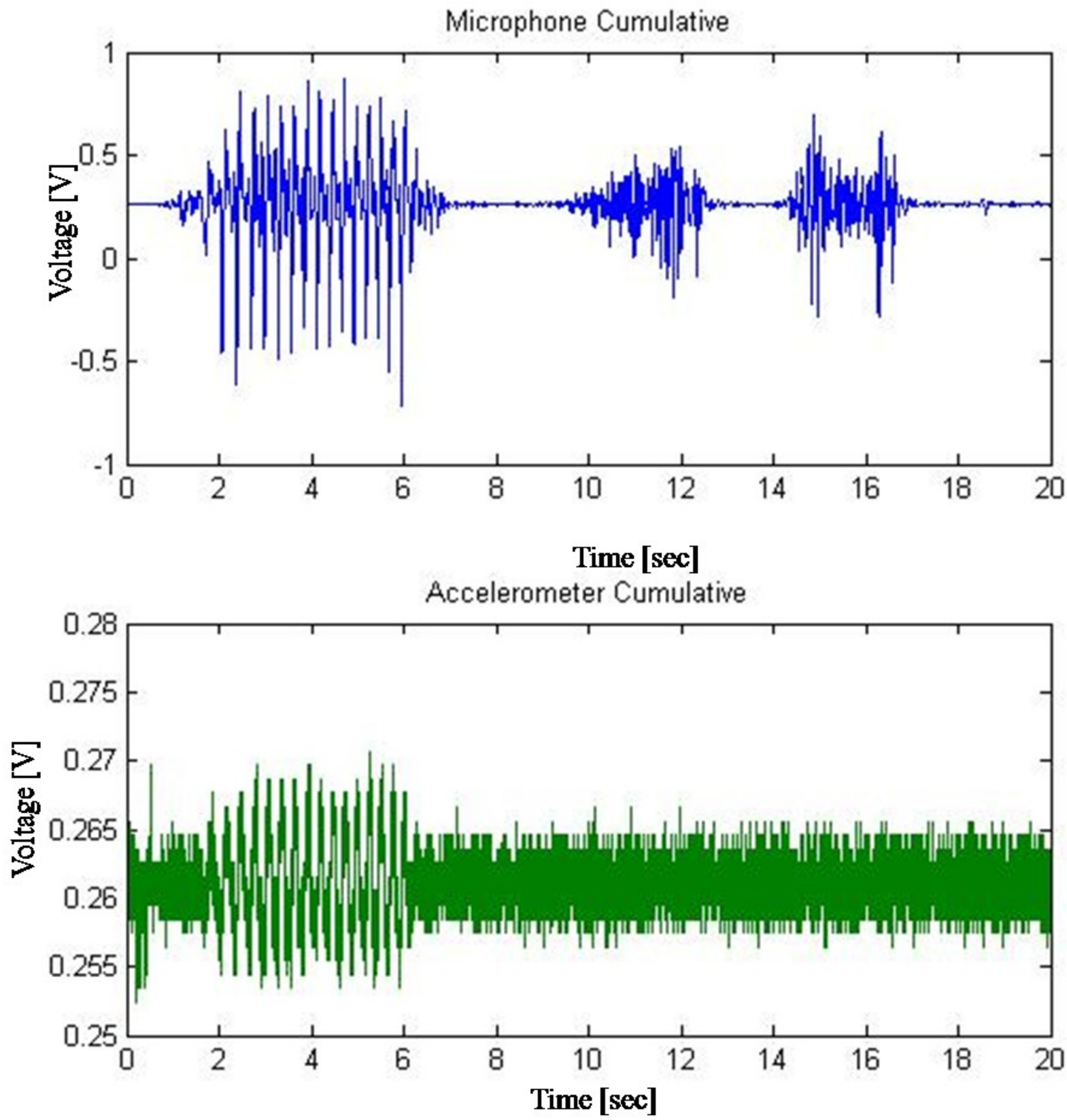


Figure 22. Partial screenshot of microphone (blue) and accelerometer (green) signals before embedding sensor. The first disturbance is from shaking the arm and second and third are from isokinetic voluntary contractions of the bicep muscle.

4.6.2 MMG Testing after embedding

After embedding the sensor in silicone, the microphone recorded a larger amplitude signal but the accelerometer signal remained approximately the same as expected. *Figure 23* shows the signals recorded from a single session with the embedded microphone/accelerometer couple taped to the bicep muscle. The first and second disturbances seen in the upper graph are from stationary voluntary contractions of the bicep. The third one is from rapid arm movement and the fourth and fifth are from arm movements, with added voluntary contractions. Lastly, there is a signal from an isokinetic voluntary contraction. From this, it is clear that the sensor is capable of recording muscle signals, but it can also be seen that motion of the sensor causes a large disturbance and must be filtered from muscle signal by some means. In the lower half of *Figure 23* it can be seen that the accelerometer is unaffected by isokinetic contractions (disturbances 1, 2 and 6 in the upper graph), but the motion has some effect on the accelerometer, although the amplitude of this signal is only slightly above the underlying noise observed in the measurement. Thus, the microphone captures the muscle action as intended but the accelerometer signal amplitude may be insufficient to use it to filter out motion artefacts from the signal. This is addressed in a later chapter.

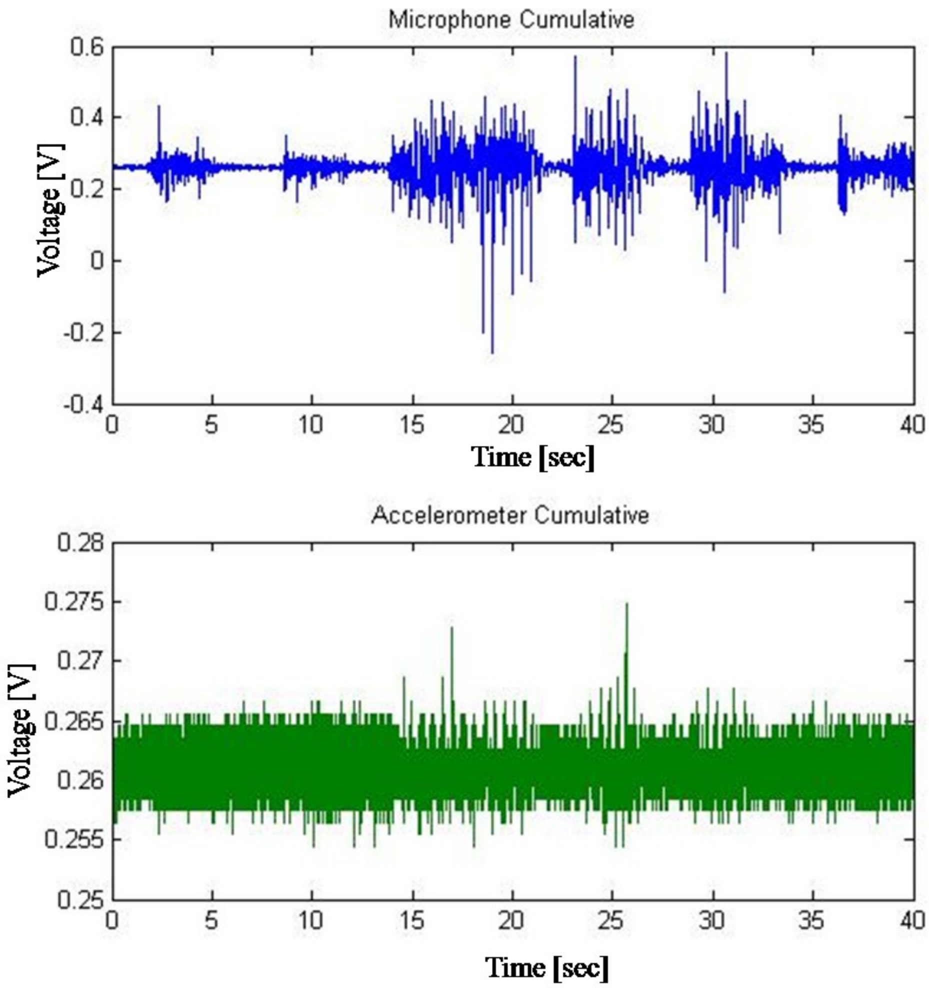


Figure 23. Partial screenshot of signals from a microphone (blue) and an accelerometer (green) after embedding. The microphone shows a large response to both muscle contractions and sensor movement but the accelerometer shows only a small response to movement.

4.7 Sensor Selection

Selecting a technology for further development was not easy as many of the investigated methods show significant potential as signals for prosthetic control. Several factors should be considered when comparing different technologies. An obviously important trait is the strength of the (voluntary) signal obtained compared to the background noise and/or interfering signal, or the signal-to-noise ratio (SNR). It should be fairly simple in construction, as a very complex sensor or sensor circuitry is more prone to failure. It is also an advantage if the use of sensor technology for a similar purpose is widespread since available information or literature will advance the development of prosthetic control using the technology. The sensor must be durable and robust enough to withstand any loads occurring during long-term usage. If activating the sensor (i.e. creating the signal) interferes with other activity this can also be a problem. This is why a simple control panel would not be an option, as it would pre-occupy at least one hand for all gait activity. The same drawback is apparent for the flexion sensor used on a toe, as bending the toes may disturb normal gait. The other sensors are all based on flexing a muscle, so it would be preferred to use a redundant muscle, e.g. a stump muscle. In *Table 1*, the sensor performance is evaluated by comparative rating in terms of signal strength, complexity, technology maturity and robustness. Positive ratings are indicated by plus signs (+) and negative ratings by minus signs (-) and one or two symbols used to differentiate between different levels of positive or negative performance.

Table 1. A comparison of different sensor technologies for user intent prosthetic control.

Attribute / Sensors		Signal strength	Complexity	Technology maturity	Robustness	Totals
EMG		--	-	++	-	++---
Force		-	+	-	-	+---
Flexion	Joint	++	+	-	-	+++--
	Shape change	-	+	-	-	+---
Inductance		--	--	-	-	------
MMG		+	+	+	+	++++

Based on this comparison it was decided to continue working with the MMG sensor, although EMG, force and flexion sensors all have potential for use as a prosthetic control input. The preliminary testing of the MMG sensor revealed great potential, but further development is necessary before a successful control system can be constructed.

5 Sensor Development

5.1 Microphone Selection

Several different commercially available miniature microphones were tested without silicon embedding to find the optimal sensor for embedding. The sensors were selected based on bandwidth, geometry, directivity and other relevant factors. The sensor circuits for each of the microphones were constructed in accordance with manufacturer specifications. *Table 2* shows an overview of the tested sensors and an evaluation of the sensors. The evaluation is based on raw sensor output amplitude from testing on a bicep muscle with a pierced silicone patch between sensor and skin; see *Figure 24*. One of the sensors did not show any signal from muscle activity, and may have been damaged during soldering. The other sensors were evaluated based on the amplitude and shape of the signal (consistency) since creating signal filters for each of the sensors and processing of all sensor data is outside the scope of this project.

Table 2. Microphones tested and compared for MMG purposes.

Microphone	Manufacturer	Bandwidth	Relative Rating
CMI-5247TF-K	CUI Inc.	70-10000 Hz	Fair
WP-23501	Knowles Acoustics	100-6000 Hz	N/A
POSM-1542-C3310-R	PUI Audio	100-7000 Hz	Poor
PVM 6027-2P423	Veco Vansonic	20-16000 Hz	Good
WM-63PRT	Knowles Acoustics	20-16000 Hz	Fair

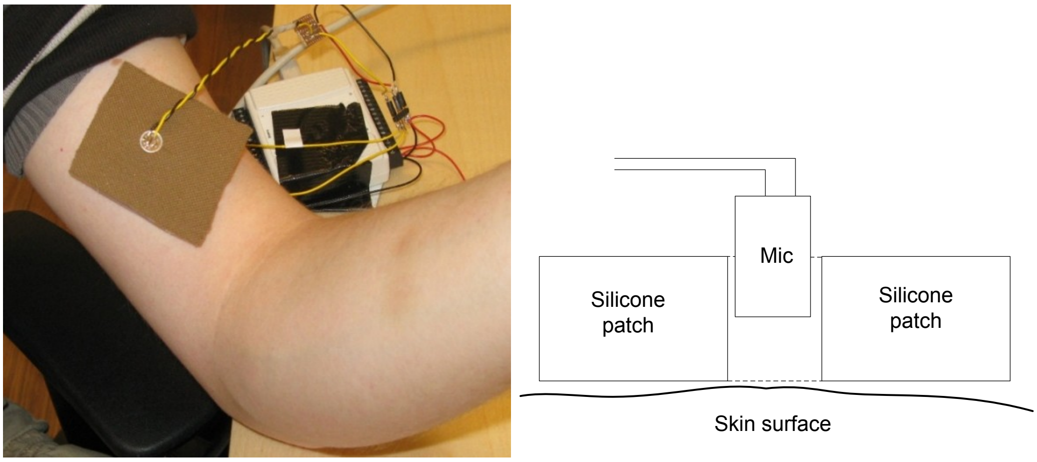


Figure 24. Testing of non-embedded microphones. Physical setup (left) and cross-sectional view (right).

The selected sensor is model PVM 6027-2P423 and it measures about Φ6.0 by 2.7 mm. The omni-directional microphone is of the electret condenser type. The simple drive circuit, soldered on a small printed circuit board (PCB), next to the microphone itself, is shown in *Figure 25*.

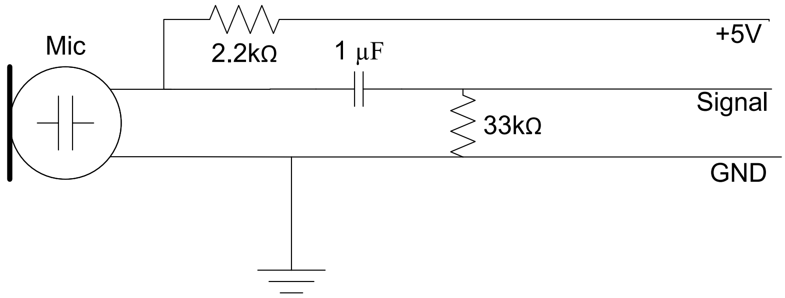


Figure 25. MMG sensor drive circuit.

5.2 Silicone Embedding

The silicone embedding and the air chamber it includes, serve to amplify the signal from muscle. For this purpose is it desirable to make the membrane from a stiff material, such that it transmits the signal from the skin to the air chamber with as little attenuation as possible. On the other hand, the material surrounding the microphone and holding it in place, relative to the skin should possess good damping abilities, so the housing of the microphone is not affected (moved) by the vibration of the skin.
The membranes are molded in silicone in a purpose made two-piece mold. The embedding of the sensor is achieved by placing it in a three-piece mold for dental silicone. The sensor is held in place in the mold by a small rod surrounding the microphone. When removed this rod also leaves the cavity for the air chamber. After embedding, the sensor membrane is glued to the embedded sensor to complete the construction.

5.3 Cancelling Microphone

The ability of the accelerometer to cancel out environmental noise is below expectations as explained in section 4.6. For this reason a new two-microphone sensor was constructed and embedded in silicone. The main microphone or the MMG microphone was adjacent to an air chamber as with the previous sensor, but the cancelling microphone was directed outwards (away from the skin) and completely molded into silicone on the opposite side of the PCB. *Figure 26* shows the signals obtained by rapid shaking of the sensor. The main microphone (blue) has a fairly symmetric amplitude of up to 1500 mV but the cancelling microphone (green) only shows an amplitude of about 2-300 mV, and the signal is not symmetric about the direct current (DC) voltage offset (at 260 mV). This may be caused by the silicone hindering the movement of the microphone membrane.

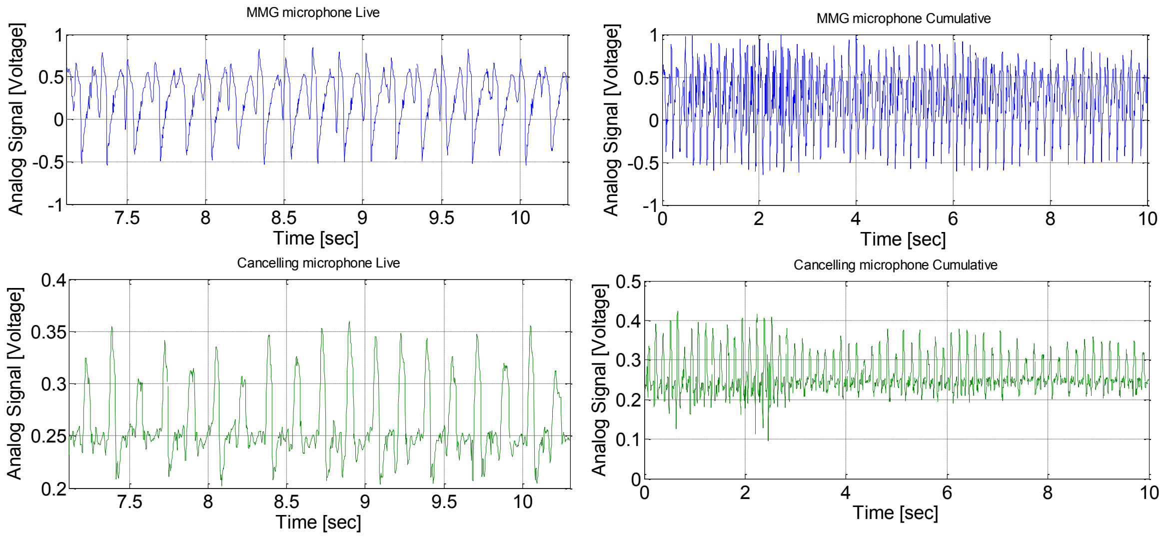


Figure 26. Rapid shaking of a two-microphone MMG sensor. The main microphone (blue) has a stronger signal and the cancelling microphone (green) has almost a single sided amplitude.

The cancelling microphone was also tested during walking, by taping it to the *tibialis anterior* muscle. The main microphone showed a much larger amplitude than the cancelling microphone in this case also; as shown in *Figure 27*. The signal from the cancelling microphone may still be suitable for filtering out movement from the main sensor, but the filtering will be more difficult, compared to two microphones with similar responses to the same movement.

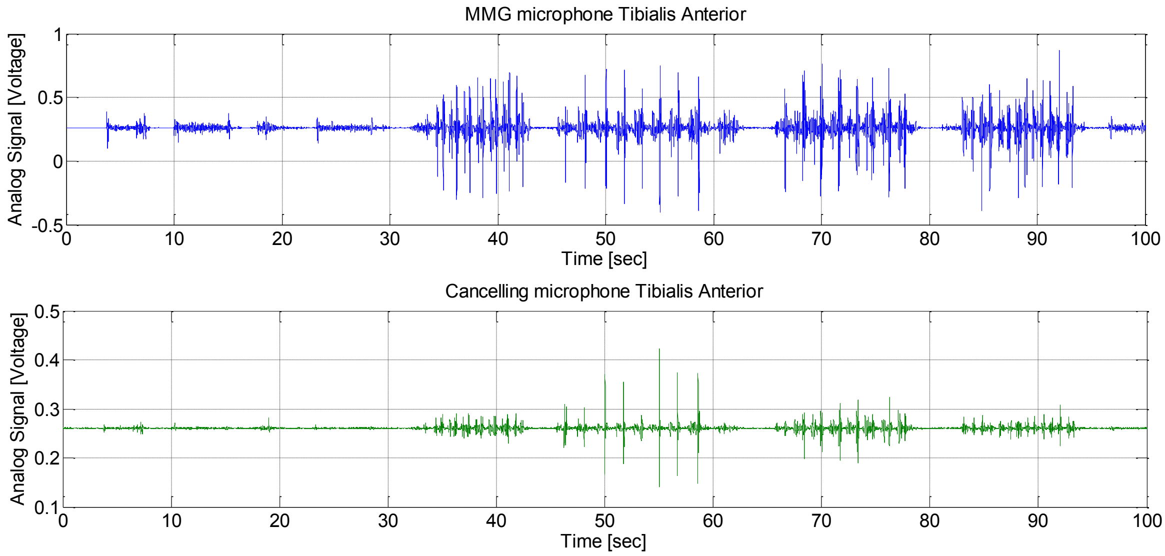


Figure 27. Two-microphone sensor tested on the Tibialis Anterior muscle. The four largest disturbances are (in order): Level walking, up-stairs walking, down stairs walking and level walking.

5.4 Noise sensitivity

A successful MMG sensor must not only be able to filter out motion artefacts, it must also be unaffected by environmental noises. To test the sensitivity of the embedded sensor to external noise, a sound level meter was used to quantify the noise from four different sources and the MMG sensor signals recorded simultaneously. The external sounds were a 1000 Hz sine wave, human rumbling, clapping of hands and an FM radio broadcast (talking). *Figure 28* shows the readings from the main MMG microphone (blue), the cancelling microphone (green) and a sound level meter (red). The sound level meter shows a clear reaction to the sound protocol, but the main MMG sensor is only affected by the 80 dB deep tone and the loud claps, and the amplitude then is less than 10 mV, compared to hundreds of mV for the muscle signal. The cancelling microphone, however shows no response above the 3-4 mV signal noise, and is thus not performing as expected. This may again be accredited to the microphone being completely encapsulated by silicone. Since the cancelling microphone was not suitable for motion or noise cancelling, it was decided to use two in-line single-microphones, and place one of the microphones distal to the muscle being measured (described in section 6.1).

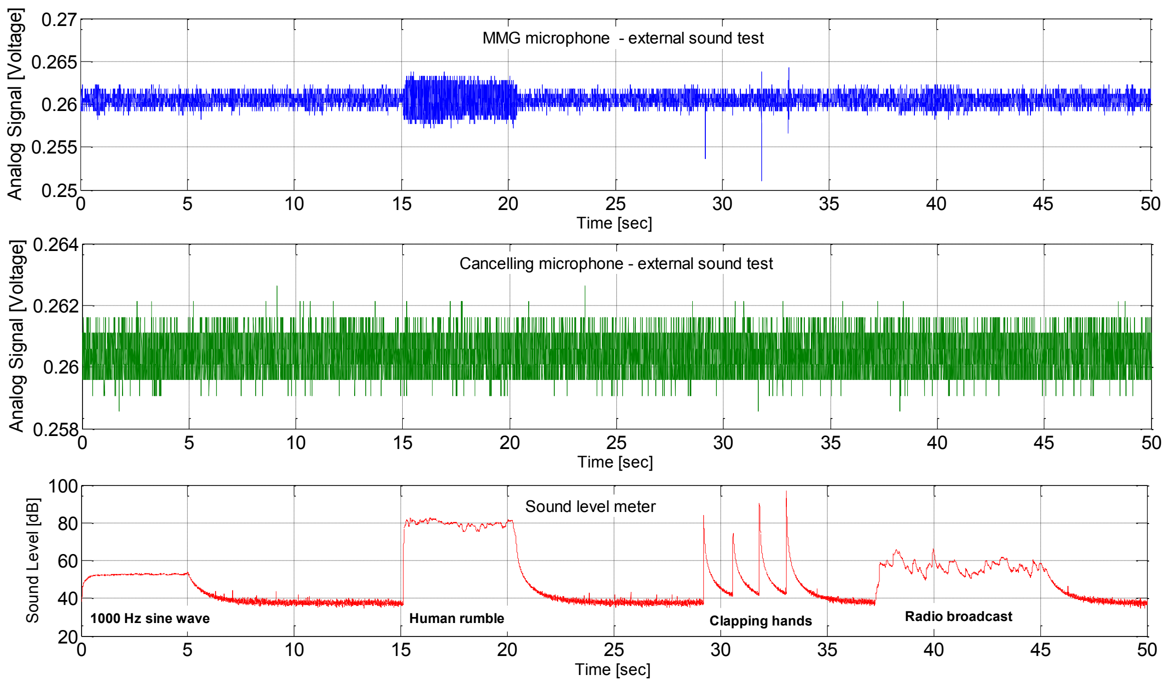


Figure 28. Noise test of two-microphone MMG sensor. A sound level meter (red) records a 55 dB, 1000 Hz sine wave, a 80 dB rumble (human), four claps, and a radio transmission at about 60 dB.

5.5 Repeatability testing

Despite the suboptimal performance of the cancelling microphone, the main microphone signal can be used for MMG in a stationary and relatively noise free environment. To test the repeatability of both the sensor and the signal it measures a test protocol was repeated daily over a period of six days. The test protocol included maximal contraction in sitting and standing position as well as a concentric contraction. The sensor was fixed in place by tape each day and this may have had an adverse effect on the repeatability. The location of the sensor was determined by skin markings. *Table 3* shows approximate amplitudes observed from each test. The main amplitude of the sitting contractions fluctuated by 50-90% in the six day period. A more consistent fastening method will presumably reduce this variability but since other signal characteristics, such as frequency and power, can also have an important role, this variability does not prevent the signals usability as a control signal. Individual or even daily calibrations can also be used to normalize the signal, provided the calibrations are simple to perform. The amplitude ranges of the standing and concentric contractions showed a much larger variation, up to 300%. It is speculated that standing contraction variability can be partly contributed to changes in posture, causing variability in muscle activation patterns. The concentric contraction signal was affected by the limb motion associated with it, and the repeatability test should therefore be repeated when sufficient motion artefact cancellation has been developed.

Table 3 Repeatability of MMG signals in different situations.

	Amplitudes [mV]		
Iteration #	Sitting Contraction	Standing contraction	Concentric contraction
1	50-60	-	140
2	40-50	25-30	90-100
3	50-60	25	60-70
4	40-50	25	80-90
5	40-50	10-20	60
6	60-75	30	50
Max variation	50-90%	50-300%	300%

5.6 Muscle selection

Selecting the right muscle(s) for prosthetic control is a difficult task but to get a general idea of what muscles should be used for normal subject testing, signals of several different leg muscles were tested. In each case a few maximal isokinetic voluntary contractions were performed and compared against background noise level and a concentric muscle action (i.e. lifting the leg to activate the muscle being testing). The results from the test session are summarized in *Table 4*:

Table 4. Signal amplitudes of several leg muscles.

Muscle	Max voluntary contraction amplitude [mV]	Concentric contraction amplitude [mV]	Noise level [mV] (i.e. muscle relaxed)
Adductor longus	50-80	100	10-20
Biceps femoris	100-150	150-400	50-70
Peroneus	100	100	10-20
Rectus femoris	60-120	120	10-20
Semitendinosus	70-100	120-150	10-20
Tibialis Anterior	50-200	200-400	50
Vastus Medialis	80-140	100-200	20-40
Gastrocnemius	100-150	120-200	20

Generally, the optimal muscle would have a large amplitude of a voluntary contraction, but a small amplitude in walking or other use. Selecting the muscle will also depend on type of amputation as parts or all of some muscles may have been amputated. Furthermore, if the control will rely on brain or neural plasticity, i.e. having the user learn to use the muscle to control the prosthesis, it may be very beneficial to select a muscle or muscles that were used for the same movement before amputation as the intended prosthesis should perform after amputation. It was therefore decided to use the signals from the *gastrocnemius* and the *tibialis anterior* as control signals for a prosthetic foot (i.e. the Proprio® foot).

6 Prototype Construction

When suitable muscles for MMG measurements have been selected, a method for securely attaching the sensors to the selected muscle sites must be found. For this purpose, it was decided to build a socket fitting a normal individual. Signals from the sensors in the socket were captured and displayed on a PC, and used to control a prototype ankle. The following sections describe the construction of these prototypes.

6.1 Socket Prototype

Since both the accelerometer/microphone couple and the two-microphone sensor had shown less than optimal performance regarding noise cancelling, it was decided to use two single microphone sensors with an air chamber in front of the microphone and a thin membrane in front of the air chamber, to capture signals from each muscle. One of the sensors is located directly on the skin and the other on the outside of the prosthetic hard socket, as shown in *Figure 29*.

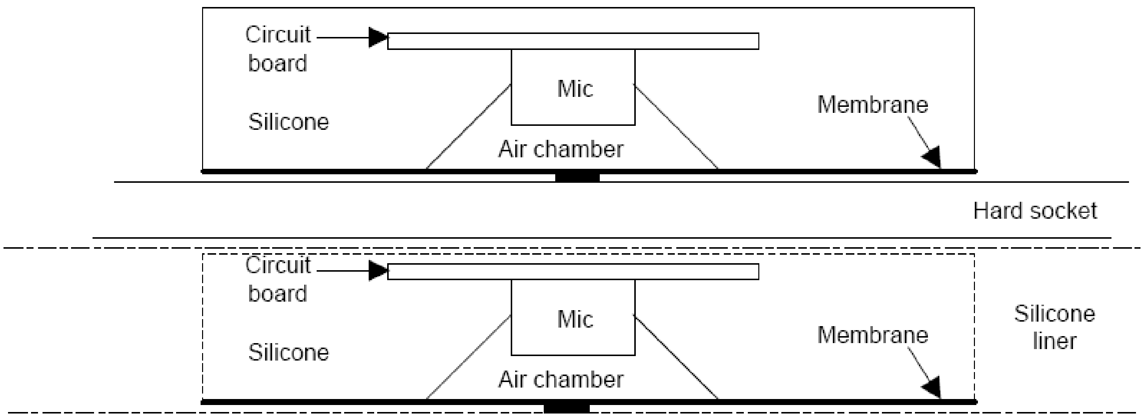


Figure 29. Two-sensor configuration. One sensor is located inside the prosthetic socket and the other on the outside of the socket.

A socket prototype for testing on a sound leg was made from a used glass-fibre prosthetic socket. Instead of a liner, two pairs of sensors were molded inside the socket using dental silicone, measuring signals from the *tibialis anterior* and medial *gastrocnemius*, respectively. The dental silicone molding also provides a good fit to the subject´s leg and should allow reasonable location accuracy between and within individual test setups. The membrane of a single sensor (grey) is visible in the prototype socket in *Figure 30*. The prototype is intended to simulate placing one sensor in the silicone liner, used inside the hard socket, by amputees, and placing a cancelling microphone on the outside of the prosthetic socket.

Figure 30. Prototype socket with sensors, for normal subject testing. The sensor (grey) is molded into the socket with dental silicone (blue).

Since both the MMG signal and the motion artefact are low frequency signals, the response of the sensors in the socket at low frequencies are of particular interest. For this reason, a comparative frequency response test of the socket sensors was conducted. A sine wave signal with varying frequency from 6 Hz to 250 Hz was played by a bass speaker (Bose Panaray MB4, frequency range 40-300 Hz ±3 dB) and the signals from the sensors of the socket, located 3 m in front of the speaker recorded. This frequency response is shown in *Figure 31*, but it should be noted that since parts of the spectrum are outside the specified range of the speaker, this frequency response should only be used for comparing the individual sensors of the socket, and not as an absolute measurement of the microphone sensitivity. It can be seen that the response of the two main microphones (red and blue) is very similar with a maximum response in the range of 45-70 Hz. The cancelling microphones are also similar to each other, and have a maximum response at 45-75 Hz, but the amplitude is approximately a quarter of the main microphone response at this frequency. This inevitably means that the cancelling microphone response to movement will be less than the main microphone response, but the same should be true for the muscle signals, and the socket must therefore be tested to judge whether this amplitude ratio is acceptable or not.

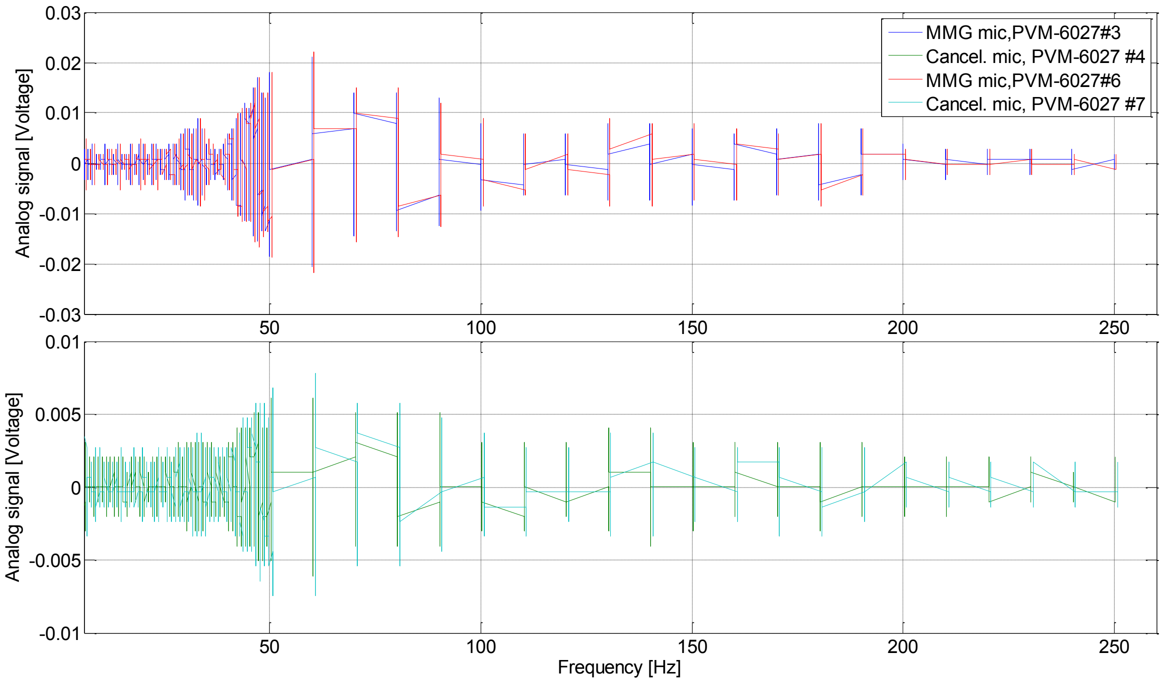


Figure 31. Comparative frequncy response of MMG and cancelling microphones of a prosthetic socket. The main microphones (MMG) have significantly larger responses than the cancelling microphones.

6.2 Ankle Prototype

In order to test the MMG sensor´s capability to control a prosthetic device, a prosthetic ankle was built from a Proprio® foot module[1] (*Figure 32*). The prototype was intended to function in a stationary setting, on a transportable wheel table or even as a part of a self-contained prosthesis for amputee testing. The Proprio foot is single axis active ankle with a battery powered electric motor actuator. The Proprio® foot does not provide propulsive power during stance phase but it improves gait by adaptation in the swing phase (off ground). The motor is controlled by an artificial intelligence processor, which was disconnected from the motor for this project. The motor was instead controlled by an ST Practispin™ L6208PD servo motor controller. This controller was set up to receive control commands from a Matlab program via serial communication. The motor was powered by a pair of 10.8 V Li-ion batteries or a stationary power supply.

Figure 32. Proprio® foot used for prototype construction

[1] http://ossur.com/?PageID=13460

6.3 Data Acquisition and Motor Control

To collect and analyze the data from the sensors in the socket and control the ankle motor a laptop PC was used as in previous testing. The Matlab program was used to capture signals from the sensors via a 12-bit NI USB-6008 AD converter. The program processes the signal and selects the appropriate motor command every 250 milliseconds. This command is sent as text via a serial communication to the motor controller board, which in turn moves the ankle motor. An overview of the system components and their connections is shown in *Figure 33*. The Matlab program has a user interface providing visual feedback by continuous real-time plotting and a logging option. A screenshot from the interface is shown in *Figure 34*.

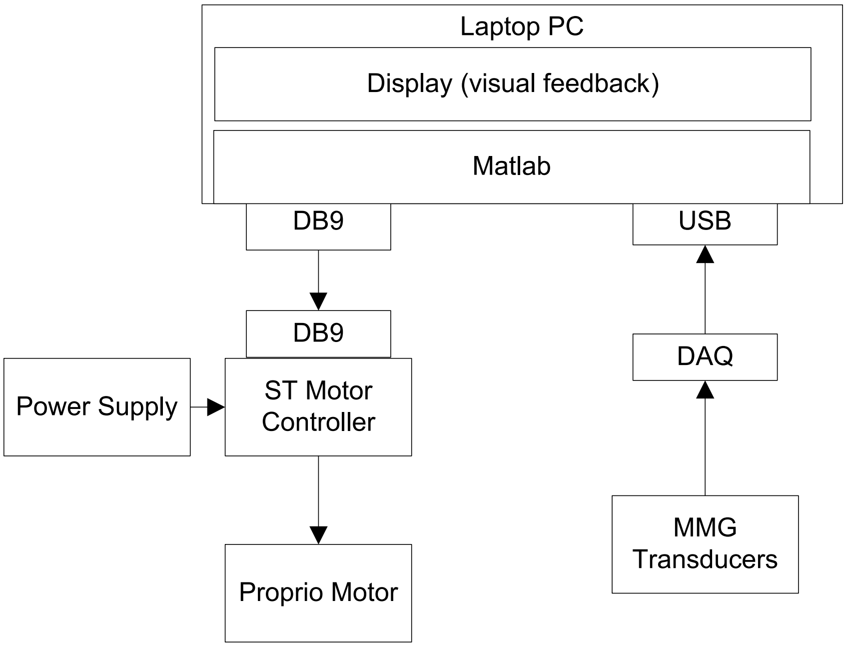


Figure 33. Schematic diagram for an MMG control prosthetic ankle.

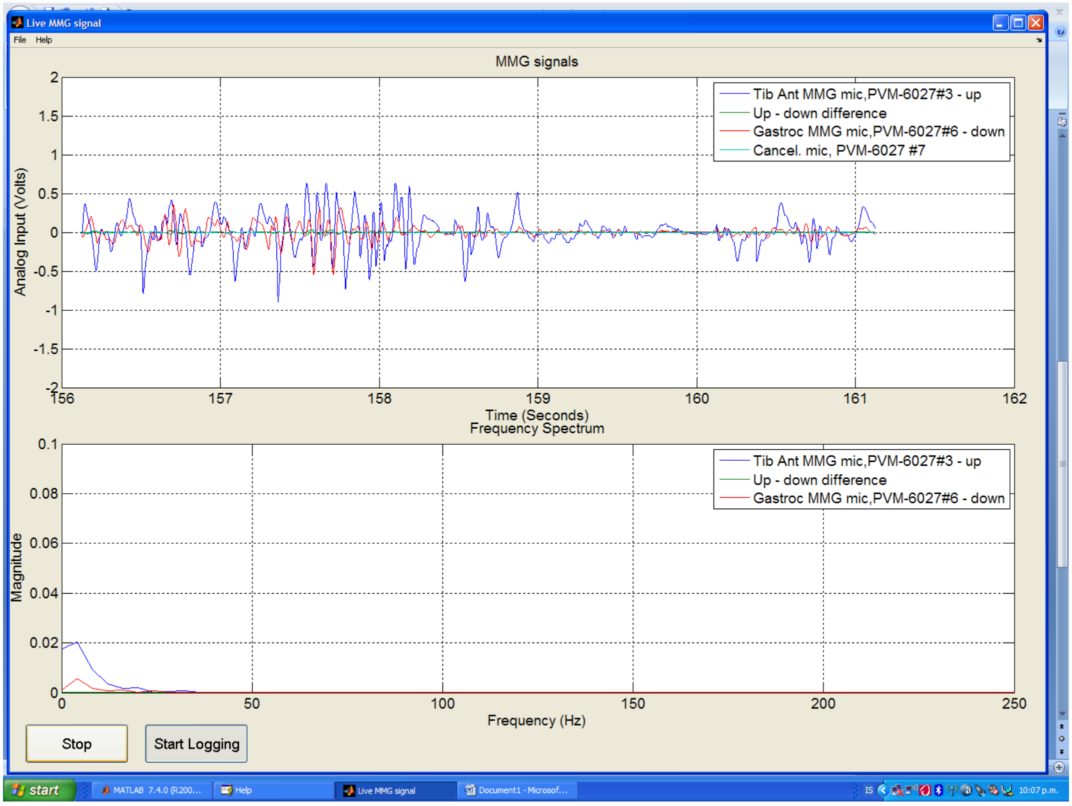


Figure 34. User interface screenshot of a real-time MMG signal plotting program and FFT analysis used for creating a signal processing method for prosthetic control.

7 Prototype Testing

A successful control system for a prosthetic ankle must of course work flawlessly in mobile and noisy settings but to aid design of a prototype capable of this, testing was divided in two parts. First, the prototype is tested in a stationary setting to develop a signal processing method to recognise muscle activation patterns without interference from motion artefact or external noise. In the second part, the signals from cancelling microphones are compared to the muscle signals and methods for filtering are introduced.

7.1 Stationary Testing

7.1.1 Signal Processing

In stationary testing, i.e. without moving the leg and with the ankle prototype fixed to a table, signals from the two main sensors (inside the socket, on both muscles) were used to control movement of the ankle. The noise cancelling microphone signals were discarded as motion artefact should be minimal in this type of testing. Several different processing methods, including moving average and spectral analysis, were tested visually with the interface described in chapter 6; an example of the signals observed is shown in *Figure 35*.

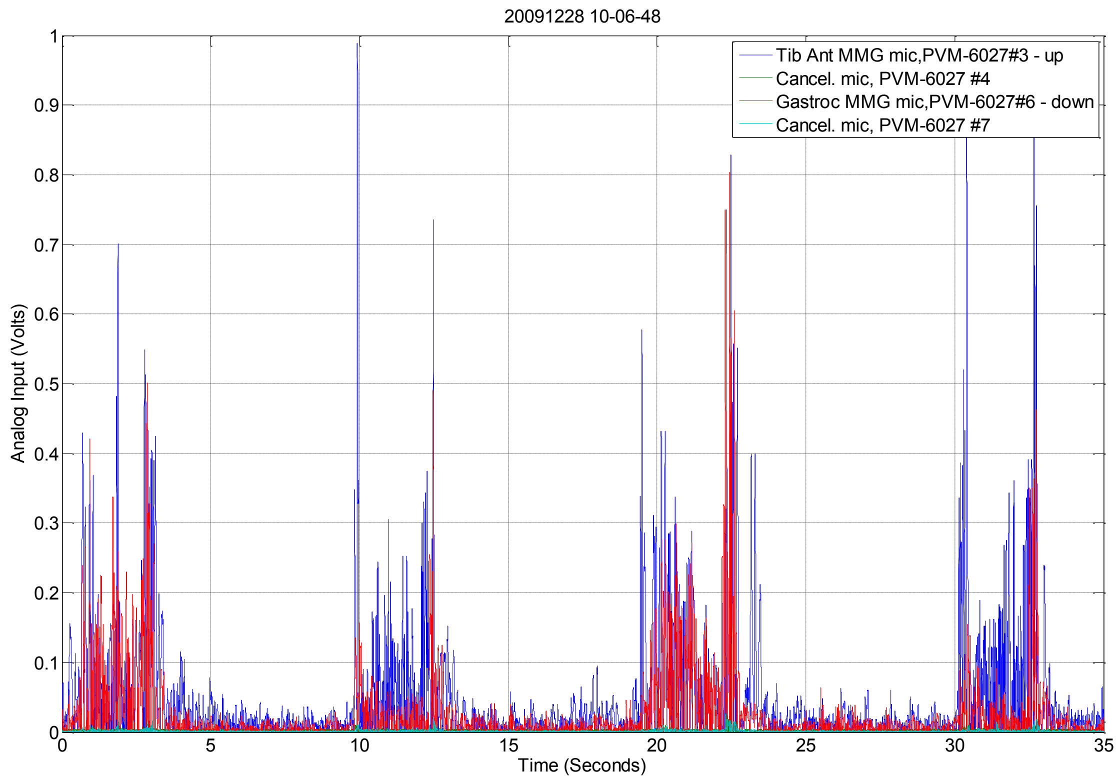


Figure 35. Example of MMG signals observed in stationary testing. The four largest amplitude periods are results of moving the foot up (first and third) and down (second and fourth) alternately.

The first and third large amplitude periods (a few seconds each) are caused by plantar-flexion (moving foot down) and the second and fourth periods are caused by dorsi-flexion (moving foot up). Different activation criteria were created and tested by comparing the prototype ankle movements to physical ankle movements. Since the *tibialis anterior* and *gastrocnemius* are antagonists (i.e. they control motion of the same joint, but in opposite directions), signal increase from both muscles was observed when moving the ankle joint. A detailed study of the lower leg muscles and their activation patters may result in "cleaner" signals but for this project, a fairly simple method of using the signal differences was used.
The DC-offset of the real-time signal was removed and the signal rectified. By finding and scaling the peaks of 250 ms "windows" of samples, suitable on/off thresholds for the difference of the signals could be determined empirically. While this reaction time is not sufficient for walking, it can give a good indication of the performance of the sensors. A descriptive pseudo-code for the resulting signal processing is as follows:

```
While program running
    For 250ms
        Signal 1 = read sample ( gastrocnemius )
        Signal 2 = read sample ( tibialis anterior )
    End

    Remove offset ( both signals )
    Rectify ( both signals )
    Scale ( both signals  )

    If      peak ( Signal 1 )  -   peak ( Signal2 )  > threshold 1
                Run motor  down
    Else-if  peak ( Signal 2 ) - peak ( Signal2 ) > threshold 2
                Run motor up
    Else
                Stop motor
end
```

After the threshold values had been selected, a training session, including visual feedback of processed signals, of less than thirty minutes preceded a series of classification tests.

In the classification test, the subject was sitting with the foot flat on the ground. The subject voiced his intention and moved the natural ankle simultaneously. The resulting movement of the prototype prosthetic ankle was compared to the natural ankle movement and voiced intention. The total number of movement instructions was 50 in each test and four consecutive tests were conducted, without removing the prototype in the 10-15 min period between tests. The test results are shown in *Table 5*. The classification accuracy ranged from 68% to 90% with an average of 83%. A better accuracy is required for prosthetic control but the test results do demonstrate the feasibility of using these sensors to capture the intent of the user.

Table 5 Classification accuracy of a stationary MMG control system.

Test#	Instructions given	Correct movements	No movement	Wrong direction movement	Wrong duration movements	Classification accuracy
1	50	45	2	2	1	90%
2	50	42	0	4	4	84%
3	50	44	2	2	2	88%
4	50	34	7	6	3	68%
Total	200	165	11	14	10	83%

7.2 Mobile Testing

7.2.1 Level ground walking

To test the suitability of the cancelling microphones signals for cancelling motion artefact, the prototype socket signals were recorded during level ground walking at a self selected speed. The results were not as expected; the amplitude of the cancelling microphone signals was almost an order of magnitude smaller than the main sensor signals, whereas the frequency response measurements in section 6.1 suggested that the cancelling microphone signals would be closer to half of the motion artefact signals of the main microphones. However, the main sensor signal contains a sum of the motion artefact and the MMG signals, but it was evident in the testing that impacts from heel strike were more prominent in the main sensor signals than in the cancelling sensor signals. *Figure 36* shows an example of the four captured signals in one of the level ground walking sessions. The cancelling microphone signals from the same test are also shown in *Figure 37* since they are not easily distinguished in the former graph. The periodic signals of about 100 mV are presumably caused by heel strike impact. These smaller signals can presumably be amplified and used to filter motion artefact but in order do that it is necessary to differentiate between the muscle signal component and the motion artefact component of the main signals. This is not easily done with signals from level ground walking so an attempt to isolate the motion artefact of the main signals was made.

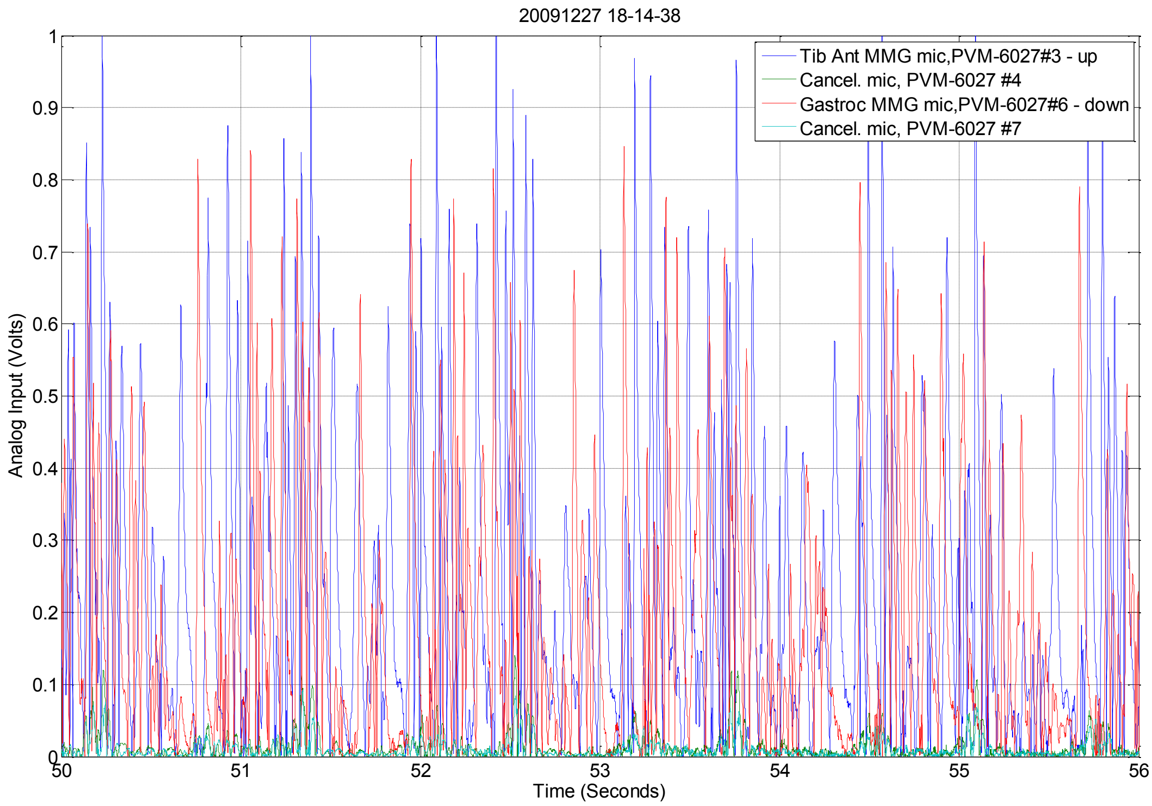


Figure 36. Signals recorded by four sensors in normal subject level ground walking. The main sensors (red and blue) have a much larger amplitude than the cancelling microphones (green and cyan).

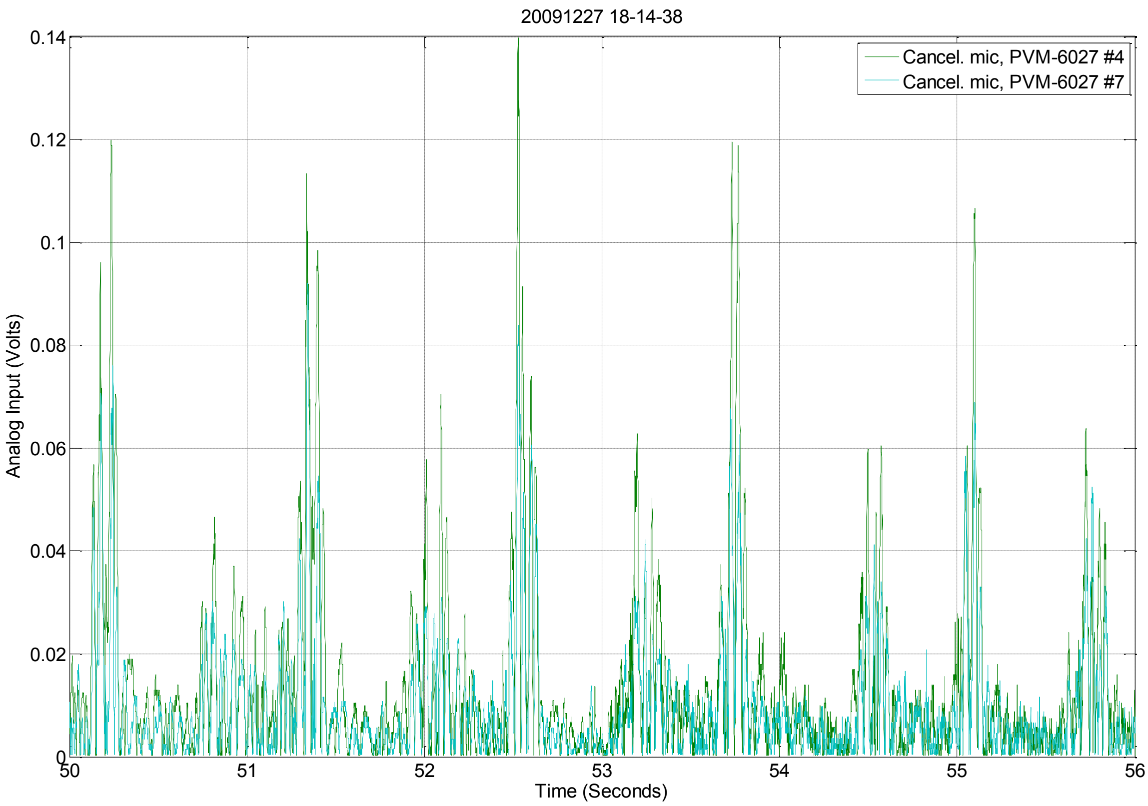


Figure 37. Cancelling microphone signals isolated from previous graph. The periodic spikes indicate heel strike and the signal may be useful for filtering motion artefact, despite the small amplitude.

7.3 Free Leg Swing Testing

To replicate the leg movements involved in normal gait without activating the *tibialis anterior* and *gastrocnemius* muscles, to which the sensors attach, the test subject was placed in a standing position, with the opposite leg in an elevated position. This allowed the free swinging of the leg being tested without ground contact. The leg was swung back and forth at a pace similar to walking speeds and an effort made not to activate the lower leg muscles. While this method cannot provide a motion artefact signal completely void of muscle signals it is a sufficient approximation for the purposes of this project. This produced an oscillatory amplitude envelope with maximum amplitude of 0.2 V, with single signal spikes reaching over 0.4 V.

The test protocol then altered to include small movements of the free-swinging foot. These movements caused large amplitude signals, compared to the motion artefact as shown in *Figure 38*, where three dorsi-flexion movements are marked with “up” and two plantar-flexion movements are marked with “down”. Since the Proprio® foot only moves the motor in the swing phase (when the foot is off the ground), this signal can be used without filtering the motion artefact from the signal, although a ground contact sensor or similar is needed to establish when the foot is off the ground. Implementing this solution is left for future researchers.

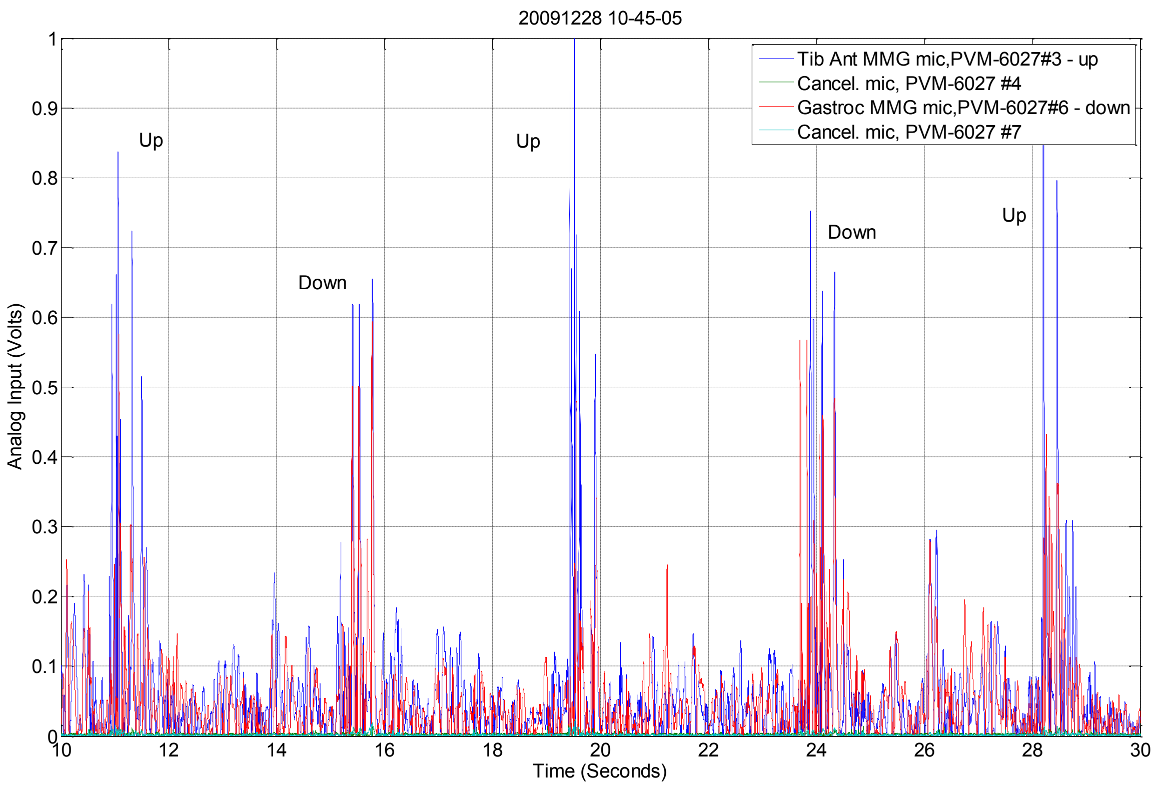


Figure 38. MMG signals during free leg swing testing. Dorsiflexion and plantarflexion are indicated by "Up" and "Down", respectively.

8 Conclusions

In this project, the following research questions were put forward:

- What type of available sensors can be used for obtaining voluntary control of lower limb prostheses?
- Can the selected technology provide sufficiently accurate and reliable information for lower limb prosthetic control?

Four different plausible methods of detecting user intent were identified and tested, all focusing on measuring muscle activity, namely, electromyography (measuring electrical potential of muscle membranes), mechanomyography (measuring muscle vibration), force sensing (force from muscle on a prosthetic socket) and using a flexion sensor for detecting muscle shape change. Mechanomyography was selected for further development, and a silicone embedded microphone sensor construction was developed, based on existing literature. An artificial socket with two sensor pairs was constructed and used to obtain muscle signals from the tibialis anterior and the gastrocnemius. The muscle signal was used to control a purposely-built prosthetic ankle prototype in a stationary setting with more than 80% classification accuracy. It was demonstrated that mechanomyography can be used to detect user intent and control a prosthetic ankle in a stationary setting but movement of the limb and sensor creates a signal with similar characteristics as the muscle signal. When captured by the sensor developed in this project, the amplitude of this unwanted signal in the swing phase of normal gait is significantly smaller than the signal from the muscles. This demonstrates the feasibility of using mechanomyography for powered prostheses that are only activated in swing phase.

Testing revealed that using the sensor pairs for cancelling motion artefact in normal gait does not have a simple straightforward solution. However, this can be addressed by enhancement of the sensor construction (to passively amplifying the muscle signal) and improving filtering techniques (to distinguish user intent from motion artefact). Filtering techniques could include frequency spectrum analysis, but since both signals are of similar frequencies, the use of other characteristics of the signals is suggested, e.g. the shape of individual oscillations or a simple pattern recognition system. It is however also critical that response of the MMG control system is consistent and predictable repeatable, in other words, a user will feel that the individual muscle contractions always result in the same response of the prostheses. By these means, it could be possible to create a prosthetic limb that functions as a natural extension of the user´s body, although the partial loss of proprioception involved in losing a limb may require some type of positional feedback for a completely natural extension of the human body.

For practical reasons, the testing in this project was mostly constrained to normal or able-bodied subject testing. This approach has both positive and negative effects on the applicability of the results for prosthetic control devices. Since the movements of the prosthetic and natural ankle can be compared, a method for controlling the prosthetic ankle like a natural ankle can be developed. On the other hand, signals from partially amputated muscles can be very different in nature, compared to normal muscles, and developing a self contained prototype for amputee testing is therefore an inevitable part of further progress of the results presented here. As discussed in section 2.1, a muscle activity based control system will inevitable lengthen the reaction time of the system, compared to normal human ankle. Modern electronics can minimize this lengthening, but the critical issue is the minimum required sampling time, before a muscle signal can be distinguished. This warrants further research into the nature of the muscle vibrations, underlying the mechanomyography technology. Although it may be possible to shorten this reaction time by e.g. reading nerve signals, this method of using the stump muscles in every step during gait may have the side effect of counteracting muscle atrophy and reduction in stump size, which is a common problem among amputees (Isakov, Burger et al. 1996).

Future research in the area of mechanomyography for control of prosthetics will hopefully lead to new and improved prosthetic products, improving the quality of life for the many people experiencing limitations in their daily lives due to the loss of a limb.

References

Aeyels, B., W. Van Petegem, J. V. Sloten, G. Van der Perre and L. Peeraer (1995). An EMG-based finite state approach for a microcomputer-controlled above-knee prosthesis.

Al-Zahrani, E., C. Gunasekaran, M. Callaghan, P. Gaydecki, D. Benitez and J. Oldham (2009). "Within-day and between-days reliability of quadriceps isometric muscle fatigue using mechanomyography on healthy subjects." Journal of Electromyography and Kinesiology **19**(4): 695-703.

Arieta, A. H., R. Katoh, H. Yokoi and Y. Wenwei (2006). "Development of a multi-DOF electromyography prosthetic system using the adaptive joint mechanism." Applied Bionics and Biomechanics **3**(2): 101-112.

Au, S., M. Berniker and H. Herr (2008). "Powered ankle-foot prosthesis to assist level-ground and stair-descent gaits." Neural Networks **21**(4): 654-666.

Barry, D. T., S. R. Geiringer and R. D. Ball (1985). "Acoustic myography: a noninvasive monitor of motor unit fatigue." Muscle & nerve **8**(3): 189-194.

Barry, D. T., J. A. Leonard Jr, A. J. Gitter and R. D. Ball (1986). "Acoustic myography as a control signal for an externally powered prosthesis." Archives of physical medicine and rehabilitation **67**(4): 267.

Coburn, J. W. (2005). Responses of mechanomyography, electromyography, and peak torque to three days of velocity-specific isokinetic training, University of Nebraska.

Courteville, A., T. Gharbi and J. Y. Cornu (1998). "MMG measurement: a high-sensitivity microphone-based sensor forclinical use." IEEE Transactions on Biomedical Engineering **45**(2): 145-150.

Cramer, J. T., T. J. Housh, J. P. Weir, G. O. Johnson, J. M. Berning, S. R. Perry and A. J. Bull (2004). "Gender, muscle, and velocity comparisons of mechanomyographic and electromyographic responses during isokinetic muscle actions." Scandinavian Journal of Medicine and Science in Sports **14**(2): 116-127.

Ebersole, K. T., K. M. O'Connor and A. P. Wier (2006). "Mechanomyographic and electromyographic responses to repeated concentric muscle actions of the quadriceps femoris." Journal of Electromyography and Kinesiology **16**(2): 149-157.

Farrell, T. R. and R. F. Weir (2007). "The optimal controller delay for myoelectric prostheses." IEEE Transactions on Neural Systems and Rehabilitation Engineering **15**(1): 111.

Garcia, G. A., F. Zaccone, R. Ruff, S. Micera, K. P. Hoffmann and P. Dario (2007). Characterization of a New Type of Dry Electrodes for Long-Term Recordings of Surface-Electromyogram.

Grass Technologies (2009). "Acoustic Myography." Retrieved Aug. 15, 2009, from http://www.grasstechnologies.com/knowledgebase/amg_demo.pdf.

Herr, H. and A. Wilkenfeld (2003). "User-adaptive control of a magnetorheological prosthetic knee." Industrial Robot: An International Journal **30**(1): 42-55.

Horn, G. W. (1972). "Electro-control: am EMG-controlled A/K prosthesis." Medical and Biological Engineering and Computing **10**(1): 61-73.

Huang, H., T. Kuiken and R. Lipschutz (2009). "A strategy for identifying locomotion modes using surface electromyography." IEEE Trans Biomed Eng **56**: 65-73.

Isakov, E., H. Burger, M. Gregoric and C. Marinccek (1996). "Stump length as related to atrophy and strength of the thigh muscles in trans-tibial amputees." Prosthetics and Orthotics International **20**(2): 96-100.

Jin, D., J. Yang, R. Zhang, R. Wang and J. Zhang (2006). "Terrain Identification for Prosthetic Knees Based on Electromyographic Signal Features." Tsinghua Science & Technology **11**(1): 74-79.

Kato, R., T. Fujita, H. Yokoi and T. Arai (2006). Adaptable EMG Prosthetic Hand using On-line Learning Method-Investigation of Mutual Adaptation between Human and Adaptable Machine.

Kondo, T., O. Amagi and T. Nozawa (2008). Proposal of anticipatory pattern recognition for EMG prosthetic hand control.

Konrad, P. (2005). "The ABC of EMG." A Practical Introduction to Kinesiological Electromyography.

Lebedev, M. A. and M. A. L. Nicolelis (2006). "Brain–machine interfaces: past, present and future." TRENDS in Neurosciences **29**(9): 536-546.

Mainardi, E. and A. Davalli (2007). Controlling a prosthetic arm with a throat microphone. Engineering in Medicine and Biology Society, 29th Annual International Conference of the IEEE.

Myers, D. R. and G. D. Moskowitz (1981). "Myoelectric pattern recognition for use in the volitional control of above-knee prostheses." IEEE Transactions on Systems, Man and Cybernetics **11**(4): 296-302.

Nashner, L. M. (1977). "Fixed patterns of rapid postural responses among leg muscles during stance." Experimental Brain Research **30**(1): 13-24.

Otto Bock (2009). "SensorHand Speed."

Peeraer, L., B. Aeyels and G. Van der Perre (1990). "Development of EMG-based mode and intent recognition algorithms for a computer-controlled above-knee prosthesis." Journal of biomedical engineering 1**2**(3): 178.

Shumway-Cook, A. and M. H. Wollacott (2007). Motor Control - Translating Research into Clinical Practice, Lippincott Williams & Wilkins.

Silva, J. (2004). Mechanomyography Sensor Design and Multisensor Fusion for Upper-Limb Prosthesis Control, University of Toronto.

Silva, J. (2007). "Measure Muscle Sounds!" Retrieved Aug. 20., 2009, from http://www.instructables.com/id/Measure-Muscle-Sounds!-Part-1:-Electronic-Board/.

Silva, J., W. Heim and T. Chau (2005). "A self-contained, mechanomyography-driven externally powered prosthesis." Archives of physical medicine and rehabilitation **86**(10): 2066-2070.

Yang, J. F. and M. Gorassini (2006). "Spinal and brain control of human walking: Implications for retraining of walking." The Neuroscientist **12**(5): 379.

Ziegler-Graham, K., E. J. MacKenzie, P. L. Ephraim, T. G. Travison and R. Brookmeyer (2008). "Estimating the prevalence of limb loss in the United States: 2005 to 2050." Archives of physical medicine and rehabilitation.