Anders Fougner

Robust, Coordinated and Proportional Myoelectric Control of Upper-Limb Prostheses

Thesis for the degree of philosophiae doctor
Trondheim, April 2013

Norwegian University of Science and Technology
Faculty of Information Technology, Mathematics and Electrical Engineering
Department of Engineering Cybernetics

NTNU
Innovation and Creativity

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Ut på tur,
aldri sur,
– med en liten kaffelaff på lur

Summary

From a prosthesis user's viewpoint there is a wide range of challenges in prosthesis research, despite the recent progression in development and manufacturing of multi-function prostheses. A small part of these challenges has been solved during the work underlying this thesis.

The scope of this thesis is to review and assess the existing methods used for proportional control, develop and demonstrate methods for artifact cancellation to increase the control reliability, design and implement a viable strategy for coordinated proportional control of multiple joints, suggest an unambiguous terminology for prosthesis control systems, and contribute to the clinical assessment of the results.

The thesis is organized as a compendium of scientific papers.

Paper A contains a pilot study of how to attenuate force induced artifacts in surface electromyography by measuring the external forces.

Paper B contains a pilot study of the adverse effects of limb position on pattern recognition based myoelectric control, hereafter called the limb position effect. Papers C, D and E contain the continuation of this project. The limb position effect was resolved by using multiple limb positions in training of the control system, and further improvements were achieved by additional use of accelerometers as a measurement of the limb position (relative to gravity). It was demonstrated that these two solutions are efficient in normally limbed subjects. Inspired by this research, further studies on prosthesis users have been reported by others.

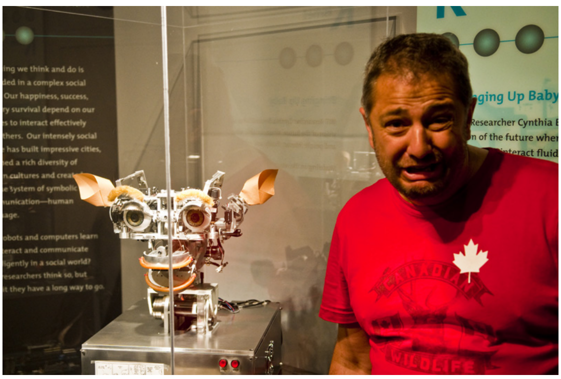
Paper F contains a comprehensive review of proportional myoelectric control of upper limb prostheses. The main findings was that the composition of the training data set and the choice of training method and optimization criterion are topics that need to be addressed in future research. This paper also contains a review of terminology in prosthesis control systems, and an unambiguous terminology has been suggested; a work that may improve communication, increase the understanding of the subject and stimulate to more structured research.

Paper G contains development and practical testing of simultaneous proportional control of two motor functions (wrist rotation and hand open/close). This required development of prosthesis guided training for proportional control, and design of a novel prosthesis socket (equivalent) for normally-limbed subjects.

This thesis has contributed towards the long-term goal of offering an intuitive and robust control system to the end users of upper limb prostheses.

Preface

This thesis is submitted in partial fulfilment of the requirements for the degree of philosophiae doctor (PhD) at the Norwegian University of Science and Technology (NTNU). The research presented here has been carried out from July 2008 to December 2012, for the most part at the Department of Engineering Cybernetics at the Norwegian University of Science and Technology, Trondheim, Norway. The research has been conducted in close cooperation with the Institute of Biomedical Engineering, University of New Brunswick (UNB), Fredericton, NB, Canada. Funding has been provided by a PhD scholarship from NTNU, as well as scholarships from the Strategic Area Medical Technology at NTNU, and The Research Council of Norway under Grant 192546 (The Leiv Eiriksson mobility programme).



Right: My supervisor Øyvind.

I would like to thank my excellent supervisor Øyvind Stavdahl, Associate Professor at Department of Engineering Cybernetics, for his enthusiasm, his huge amount of ambitious and freely shared ideas, and for his humour. I am happy to state that my supervisor's largely positive feedback has never let me strain into the right half plane, nor put me into oscillations, but rather pushed me efficiently in the right direction (of the PhD degree).

Øyvind introduced me to his good friend Peter Joseph Kyberd, Professor at the Institute of Biomedical Engineering, who has been my co-supervisor. I had the privilege to visit him and his colleagues at UNB for one year (2009–2010). I would also like to thank Prof. Kevin Englehart, Prof. Emeritus Phil Parker, Dr. Yves Losier, and Erik Scheme, all at UNB, as well as Dr. Adrian D. C. Chan at Carleton University in Ottawa, for a very fruitful cooperation initiated while I visited them. I learned a lot about biomedical intrumentation from Prof. Emeritus Dennis Lovely during the PhD course at UNB, and I enjoyed the cooperation and discussions with the PhD students at the Institute. The awesome rock & ice climbing community in Fredericton does also deserve to be mentioned. I am very grateful for their hospitality and it was a great experience. I certainly learned a lot about upper-limb prostheses, Canadian culture and rock climbing during one year at UNB.

My co-supervisor Peter, enthusiastically skiing.

I have had a great time working with my colleagues, co-workers, office mates and friends at the Department of Engineering Cybernetics. Particularly I have appreciated the discussions with Parsa Rahmanpour, Johannes Tjønnås, Arnfinn Aas Eielsen, Øyvind Stamnes, Pål Liljebäck, Martin Korsgaard, Milan Milovanovic, Espen Hauge and Christian Holden. I want to thank especially Christian for valuable feedback on LaTeX typesetting and graphics in my publications and thesis.

In parts of my PhD study I have worked closely with the students I have been co-supervising. Especially Marthe Sæther, Per Ferdinand Bach, Jørn Bersvendsen and Ådne S. Linnerud have been involved in work closely related to the material published in this thesis.

Tomm Kristensen and Bjørn L. Lien of Norsk Teknisk Ortopedi AS, Ottestad, Norway, and Hans Petter Aursand of Trondheim Ortopediske Verksted, St. Olav's Hospital, University Hospital of Trondheim, Norway, are acknowledged for their invaluable contributions to design and production of a prosthesis socket for normally-limbed subjects.

The administrative and technical staff at the Department of Engineering Cybernetics have been very helpful; especially Terje Haugen and Per-Inge Snildal at the mechanical workshop have contributed with customization of the prosthesis socket, and Stefano Bertelli has assisted with the prosthesis interface.

I would like to thank my friends for regularly bringing me out of the laboratory. As illustrated in Fig. A, skiing and mountaineering has offered motivation between long days at work.

Finally, this thesis would never have taken form without the support from my parents, Reidun and Kristian, my sister, Stine, my two nieces, Solveig and Ragne, and the rest of my family.

Trondheim, December 2012

Anders Fougner

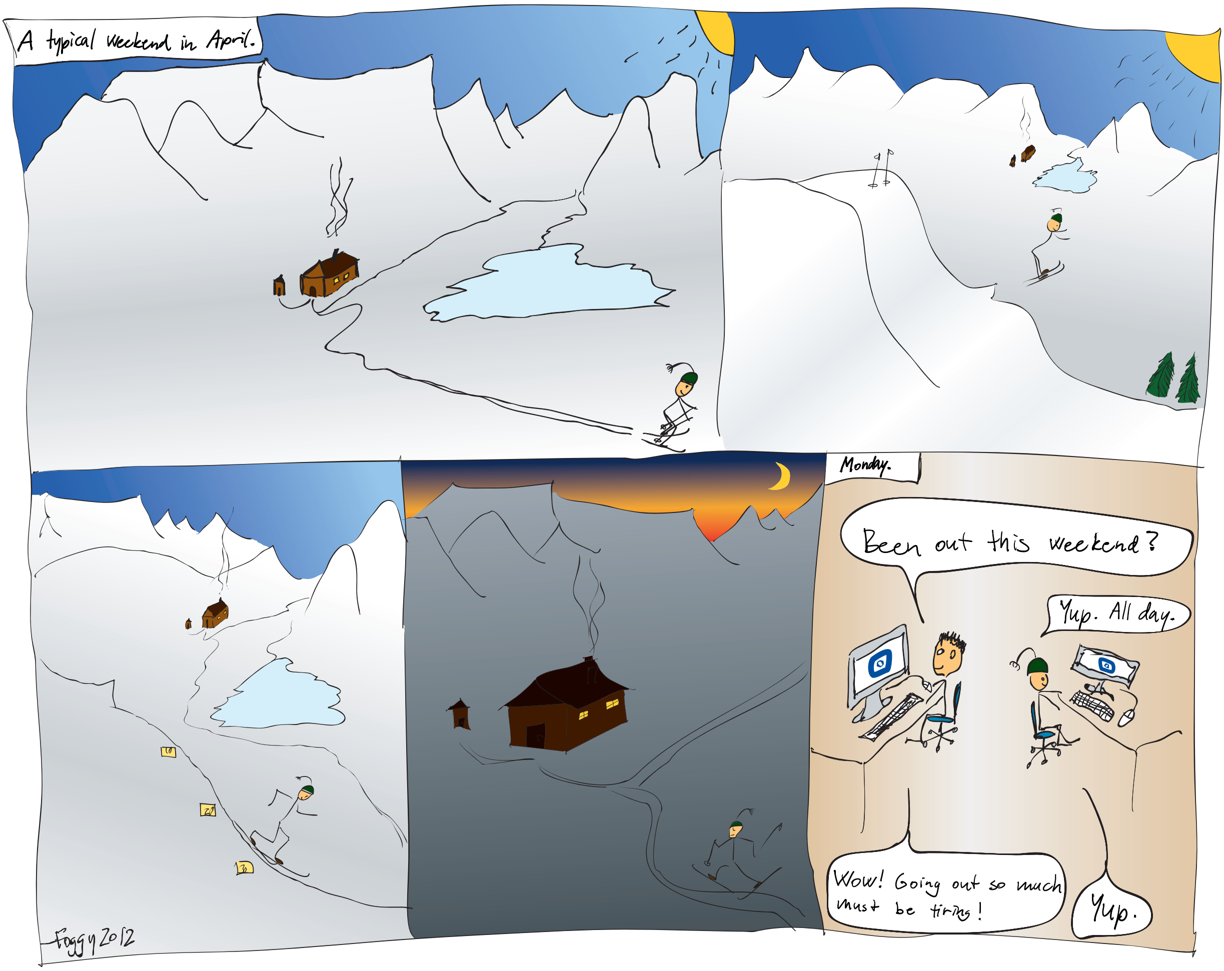


Fig. A: Working Environment in Trondheim and at the Department of Engineering Cybernetics.

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Nomenclature

PD	The image is in the public domain due to its age.
CC BY-NC-SA	The figure is licensed under a Creative Commons BY-NC-SA license. See http://creativecommons.org/licenses/by-nc-sa/3.0/.
ECG	Electrocardiography (or electrocardiogram); the recording of electrical activity of the heart over a period of time.
EMG	Electromyography (or electromyogram); the recording of the extracellular field potentials produced by muscles.
ICF	International Classification of Functioning, Disability and Health, a framework for health and disability defined by World Health Organization (2002).
LDA	Linear discriminant analysis; a method commonly used in statistics, pattern recognition/classification and machine learning.
MES	Myoelectric signal; the electrical activity produced by a contracting muscle.
PGT	Prosthesis guided training; a system training method. The prosthesis is moving while the user follows the motions with the phantom limb.
RMS	Root-mean square; a statistical measure of the magnitude of a variation.
SHAP	Southampton Hand Assessment Procedure (Light et al. 2002; Kyberd et al. 2009).
TD	Terminal device; in upper-limb prosthetics typically a hook or a hand.
TD (features)	Time-domain features; a group of signal features (Hudgins et al. 1993) from the time domain.
ULPOM	Upper Limb Prosthetic Outcome Measures group (Hill et al. 2009).

Chapter 1

Scope and Contributions

1.1 Thesis Outline

The thesis is a paper compendium and is organized as follows:

This Chapter presents the scope of the thesis and lists the publications and contributions in the present work.

Chapter 2 Introduction introduces the reader to upper-limb prostheses and myoelectric control, describes the state of the art in control of upper-limb prostheses and presents the challenges in assessment and use of a modern multifunction upper-limb prosthesis.

Chapter 3 Discussion puts the contributions and publications into context and discusses the strengths and weaknesses of the present work.

Chapter 4 Concluding Remarks summarizes the work, defines relevant topics for future work and concludes the thesis.

Chapter 5 Original Publications contains six published papers in facsimile, as well as one submitted journal paper manuscript.

1.2 Scope of the Thesis

The long-term aim of the work is to offer an intuitive and robust control system to the end users of upper limb prostheses. This thesis contributes in the development towards that goal by having the following scope:

- *To suggest an unambiguous terminology for prosthesis control systems,*
- *review and assess the existing methods used for proportional control,*
- *develop and demonstrate methods for artifact cancellation to increase the control reliability,*
- *design and implement a viable strategy for coordinated (simultaneous) proportional control of multiple joints,*
- *contribute to the clinical assessment of the results.*

1.3 List of Publications

The work underlying this thesis has produced the following publications (ordered by publication type, and chronologically numbered):

Journal Papers

Paper C: A. Fougner, E. Scheme, A. D. C. Chan, K. Englehart, and Ø. Stavdahl, "Resolving the Limb Position Effect in Myoelectric Pattern Recognition", published in the *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 19, no. 6, pp. 644–651, Dec 2011.
Pubmed ID: 21846608. DOI: 10.1109/TNSRE.2011.2163529.

Paper F: A. Fougner, Ø. Stavdahl, P. J. Kyberd, Y. G. Losier, and P. A. Parker, "Control of Upper Limb Prostheses: Terminology and Proportional Myoelectric Control – A Review", published in the *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 20, no. 5, pp. 663–677, Sep 2012.
Pubmed ID: 22665514. DOI: 10.1109/TNSRE.2012.2196711.

Paper G: A. Fougner, Ø. Stavdahl, and P. J. Kyberd, "System Training and Assessment in Simultaneous Proportional Myoelectric Prosthesis Control", submitted to the *IEEE Transactions on Biomedical Engineering*. Date of first submission: 20 December 2012, under the title "Simultaneous Proportional Control of Dualfunction Upper Limb Prostheses". Date of second submission: 7 March 2013.

Conference Papers

Paper A: A. Fougner, M. Sæther, Ø. Stavdahl, P. J. Kyberd, and J. Blum, "Cancellation of Force Induced Artifacts in Surface EMG using FSR Measurements", published in the *Conference Proceedings of the Myoelectric Controls Symposium* (MEC 2008), Fredericton, NB, Canada, Aug 2008.

Paper B: E. Scheme, A. Fougner, Ø. Stavdahl, A. D. C. Chan, and K. Englehart, "Examining the adverse effects of limb position on pattern recognition based myoelectric control", published in the *Conference Proceedings of the IEEE Engineering in Medicine and Biology Society* (EMBC 2010), Buenos Aires, Argentina, Aug 2010.
Pubmed ID: 21097173. DOI: 10.1109/IEMBS.2010.5627638.

Paper D: A. Fougner, E. Scheme, A. D. C. Chan, K. Englehart, and Ø. Stavdahl, "A multimodal approach for hand motion classification using surface EMG and accelerometers", published in the *Conference Proceedings of the IEEE Engineering in Medicine and Biology Society* (EMBC 2011), Boston, MA, USA, Sep 2011.
Pubmed ID: 22255277. DOI: 10.1109/IEMBS.2011.6091054.

Paper E: A. Fougner, E. Scheme, A. D. C. Chan, K. Englehart, and Ø. Stavdahl, "Resolving The Limb Position Effect", published in the *Conference Proceedings of the Myoelectric Controls Symposium* (MEC 2011), Fredericton, NB, Canada, Aug 2011.

Other Publications

- Ø. Stavdahl, A. Fougner, and P. J. Kyberd, "Simultaneous Proportional Control of Multiple Functions in Upper-Limb Prostheses" [Abstract], presented at the *World Congress of the International Society for Prosthetics and Orthotics* (ISPO 2007), Vancouver, BC, Canada, May 2007.
- Ø. Stavdahl, A. Fougner, and P. J. Kyberd, "Measurement of Bio-Signals" [Patent], Patent no./License no.: US prov. 61/087480, UK0814533.6, NTNU, Trondheim, Norway, August 2008.
- A. Fougner, Ø. Stavdahl, and P. J. Kyberd, "EMG Feature Selection for Simultaneous Proportional Control of Multifunctional Upper-Limb Prostheses" [Poster] presented at the *World Congress of the International Society for Prosthetics and Orthotics* (ISPO 2010), Leipzig, Germany, May 2010.
- Y. Kerlefsen, A. Fougner, and Ø. Stavdahl, "Smarter prostheses using iPhone technology" [Newspaper], published in *Gemini*, No. 4, pp. 6–7, March 2010.
- S. Ekiz, Ø. Stavdahl, and A. Fougner, "Newton: Protese og epost" [Television], in *NRK1/NRK Super*, Norwegian Broadcasting Corporation, 9 September 2012.

1.4 List of Contributions

The publications in this thesis have produced the following contributions (presented chronologically):

Paper A: Proposed to reduce force induced artifacts in surface electromyography (EMG) by measuring the external forces acting on the electrodes. The paper presents the pilot study. Further development of the work has been patented and published by Stavdahl et al. (2011); Stavdahl et al. (2012); Stavdahl et al. (to be submitted).

Paper B: Pilot study. Discovered the adverse effects of limb position on pattern recognition based myoelectric control (hereafter called the Limb Position effect). Initial demonstration of methods to overcome the problem. Further work has been published in Papers C, D and E.

Paper C:

1. Continuation of the Limb Position effect study from Paper B, on more subjects and with an improved protocol.
2. Proposed the use of multiple limb positions in the system training to overcome the limb position effect.

3. Proposed the additional use of inertial sensors (accelerometers) as a measurement of the limb position, to improve the system performance.
4. Demonstrated that the proposed solutions resolve the Limb Position effect on normally limbed subjects. Inspired by this project, further studies on prosthesis users have been reported (L. Chen et al. 2011; Jiang et al. 2012b).

Paper D: Investigated the efficacy of accelerometers in comparison to adding electrode sites, in order to overcome the Limb Position effect. It was found that if one wants to improve a "contemporary" two-site myoelectric control system, it is more advantageous to add an accelerometer affixed to the forearm rather than increase the number of electromygraphy (EMG) channels. This study complements Paper C and therefore overlaps in some parts.

Paper E: Examined the generalizability of the training set as a function of the number of limb positions in the set. It was found that most of the improvement is achieved already when increasing from one to two training positions. This study complements the results reported in Papers C and D and therefore overlaps in some parts.

Paper F:

1. A comprehensive review of terminology in prosthesis control systems. The suggested terminology may stimulate the communication between researchers, clinicians, users and other people involved in prosthetics. Simultaneously it may improve the understanding of the subject and stimulate to more structured research.
2. A comprehensive review of proportional myoelectric control of upper limb prostheses. It was discovered that the composition of the training data set and the choice of training method and optimization criterion are topics that need to be addressed in future research.

Paper G:

1. Development of *prosthesis guided system training* for proportional myoelectric control.
2. Development of a "prosthesis socket equivalent" for practical testing of prosthesis control methods on normally-limbed subjects. The socket restricts the hand, wrist and finger joints, resulting in near-isometric muscle contractions.
3. Development and practical testing of simultaneous proportional control of two motor functions (wrist rotation and hand open/close) in upper limb prostheses.

1.5 The Author's Individual Contributions in Co-authorships

Table 1.1 presents the author's contributions to each of the selected papers.

Table 1.1: Contributions to Papers A–G.

Scale:	
Has essentially done all the work independently	(90-100%)
Has done most of the work	(70-90%)
Has contributed considerably	(40-60%)
Has contributed to the collaboration	(10-30%)
No or little contribution	(0-10%)

	Paper						
Contribution	**A**	**B**	**C**	**D**	**E**	**F**	**G**
Formulation/identification of the scientific problem	(10-30%)	(70-90%)	(70-90%)	(70-90%)	(70-90%)	(70-90%)	(70-90%)
Planning of experiments, data collection or literature review	(10-30%)	(40-60%)	(40-60%)	(40-60%)	(40-60%)	(70-90%)	(70-90%)
Design, development and implementation of methodology	(10-30%)	(0-10%)	(40-60%)	(40-60%)	(40-60%)	(70-90%)	(40-60%)
Data collection and experimental work	(0-10%)	(70-90%)	(90-100%)	(90-100%)	(90-100%)	(90-100%)	(90-100%)
Interpretation of results	(40-60%)	(10-30%)	(70-90%)	(70-90%)	(70-90%)	(70-90%)	(70-90%)
Writing of the first draft of the manuscript	(70-90%)	(0-10%)	(70-90%)	(70-90%)	(70-90%)	(70-90%)	(70-90%)
Finalization of the manuscript and submission	(40-60%)	(10-30%)	(70-90%)	(70-90%)	(70-90%)	(70-90%)	(70-90%)

Chapter 2

Introduction

This chapter introduces the reader to upper-limb prostheses and myoelectric control, describes the state of the art in control of upper-limb prostheses and presents the challenges in assessment and use of a modern multifunction upper-limb prosthesis.

2.1 Background and Motivation

Humans with amputations or congenital absence exist in every country. Nobody has a complete overview of the amount of people with amputations or congenital absence, or the need for prostheses. Some researchers have tried to estimate the incidence rates and the number of prosthesis users in parts of the world. Here are some examples for upper limb prostheses:

- In Norway, the prevalence of adult acquired major upper limb amputation (through or proximal to the wrist) has been estimated to 11.6 per 100,000 adults (n=416) by Østlie et al. (2011). Most of these had used prostheses.
- Approximately 100,000 people in the USA have a major upper-limb loss, 57% of them being transradial amputees (Esquenazi et al. 1996; Ziegler-Graham et al. 2008; Merrill et al. 2011). About 80% of these use a prosthesis (Biddiss et al. 2007), and roughly 30–50% use myoelectric controlled devices (Kyberd et al. 2011; Merrill et al. 2011).

There are six major manufacturers of myoelectric upper limb prostheses (Liberating Technologies, Inc.; Motion Control, Inc.; Otto Bock GmbH; RSL Steeper; Shanghai Kesheng Prosthese Co., Ltd.; Touch Bionics, Inc.). Some of them use expressions like "mind-controlled" or "bionic" to describe their products (Otto Bock GmbH; Touch Bionics, Inc.; RSL Steeper), and these terms have in occasions been repeated in newspapers and television. There are also videos of high-level amputees demonstrating amazingly good control with their arm prostheses. Thus, people commonly believe that prosthesis users today easily can control their artificial hand through implanted sensors in their brain, nerves or muscles. On the other hand, even though the newest multifunction upper-limb prostheses have an impressive design and a long list of grip patterns to select from, none of the control methods are yet able to offer an intuitive interface to the user. There is indeed a lot of research going on in the fields of brain-computer interfaces and neuroscience, but none of this technology is yet available to the prosthesis users. Commercial myoelectric upper limb prostheses are still using a couple of surface electromyographic sensors placed on the inside of the prosthesis socket, and the control systems are often not intuitive at all – at least not for multifunction devices.

The researchers, the clinicians, the developers and the prosthesis users themselves are all aware of the need for a better and more intuitive way of controlling the upper-limb prostheses. Several new multifunction hands have been introduced during the last decade, and at the same time the upper-limb prosthetics research has seen a remarkable increase in the number of publications.

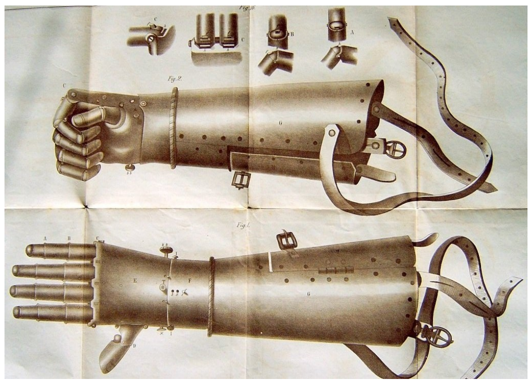
An efficient way to get new and better ideas is to look into the history and learn from experiences and thoughts that researchers of the past have found worthy of being written down and read by others. Therefore, the next few sections of this thesis contain a brief look into the history of upper-limb prostheses, accompanied by a review of proportional myoelectric control in Part III of Paper F. Both of these reviews are mainly focused on transradial prostheses, but some of the methods may also be used in higher-level devices or lower-limb prostheses.

2.2 Upper-Limb Prostheses

2.2.1 Body-Powered Upper-Limb Prostheses

The history of amputations and prostheses from the early days until 1975 has been described by VanDerwerker Jr. (1976), Putti (1925, 2005) and Norton (2007). Putti's famous example is the story of the knight, poet and adventurer Gottfried “Götz” von Berlichingen, who lost his hand in a battle in 1504. Technical expertise of workshops in the nearby cities made him a mechanical replacement hand of iron (see Fig 2.1), and at least three versions of this hand are known. Presumably it was used with success in battles. In those situations, one important property of the prosthetic hand was actually that it looked scary and that it was more robust than a healthy limb. The autobiography of Götz made the basis for one of Goethe's most famous plays, approximately 270 years later (Goethe 1848).

(a) Götz von Berlichingen



(b) Götz von Berlichingen's prosthetic hand

Figure 2.1: Franconian knight Götz von Berlichingen and a painting of his Iron Hand from circa 1509. Image sources: Putti (1925); Wikipedia (2012). Both images are in the public domain due to their age (hereafter represented by a PD symbol).

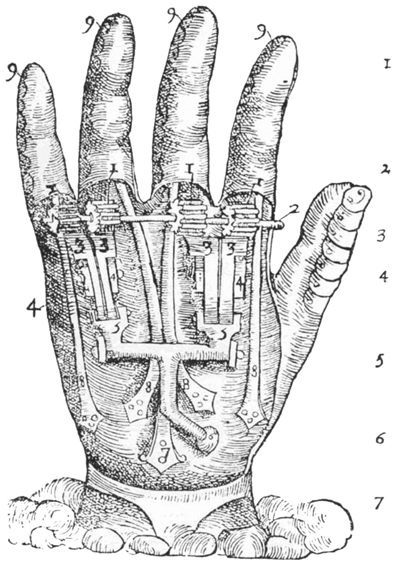
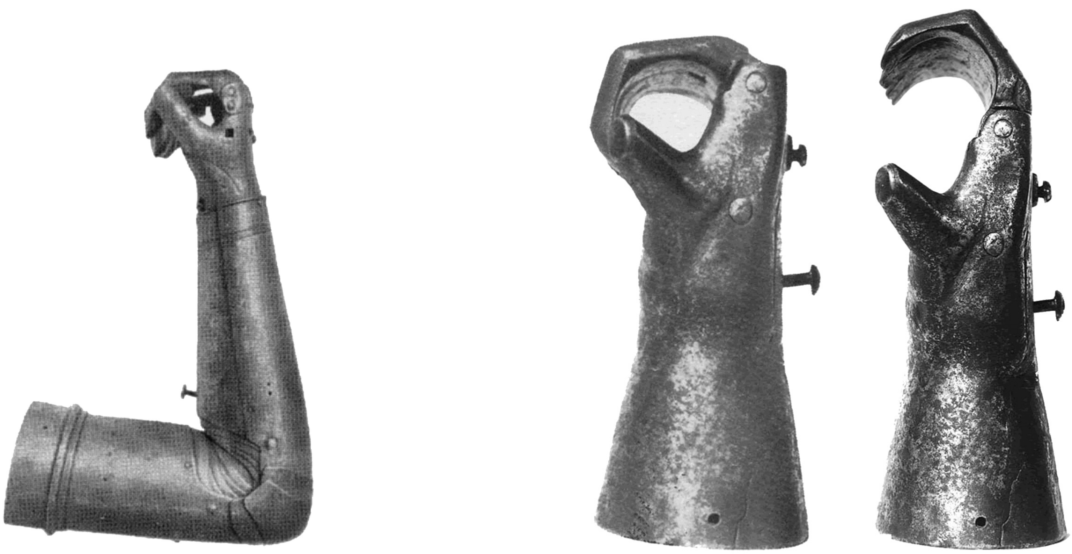


Figure 2.2: Demonstration of the mechanism in the “petit Lorrain” hand (16^{th} century). Image source: Putti (1925)[PD].



(a) (b)

Figure 2.3: Arms and hands (15^{th}–16^{th} century) from the “Stibbert” museum in Florence, Italy. Image source: Putti (1925)[PD].

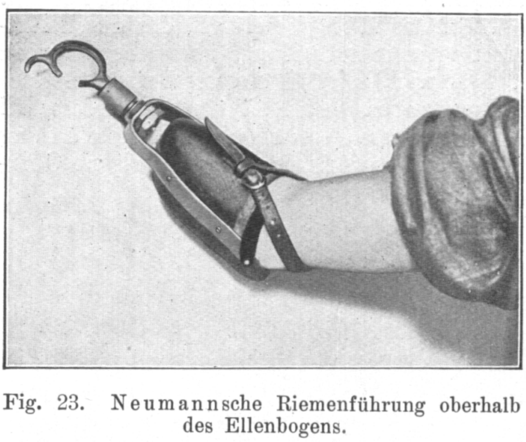
Also described by Putti are the "petit Lorrain" hand (Fig. 2.2) and the "Stibbert" hands and arms (Fig. 2.3). All of these hands from the 15^{th}–16^{th} century were inspired by the body armour used in battle at the time. They were designed with *function* and *robustness* as the main criteria, rather than aesthetics. Several of these early designs thus had joints that could be locked by a spring ratchet mechanism, through a metal lever operated by the other hand.

(a) (b)

Figure 2.4: Artificial left arm, Europe, 1850–1910. Image source: Science & Society Picture Library, British Science Museum (2012). Used with permission.

Another interesting design is the one of Fig. 2.4 from the end of the 19^{th} century. "The elbow joint can be moved by releasing a spring, whereas the top joint of the wrist allows a degree of rotation and an up-and-down motion. The fingers can also curl up and straighten out." (British Science Museum 2012). It has similar mechanisms to the older hands, but it is more lightweight, has more degrees of freedom and has a leather socket.

The next important steps in the development of upper limb prostheses have been described by Kuniholm (2010) and consist of the hook design (examples in Fig. 2.5), body-powered actuation by Selpho (1857) and Reichenbach (1865) (Fig. 2.6), and the split-hook design invented by Dorrance (1912) (Fig. 2.7a).



(a)

(b)

Figure 2.5: Passive hooks and shoulder harness by Weimar. Image source: Lange (1922)PD.

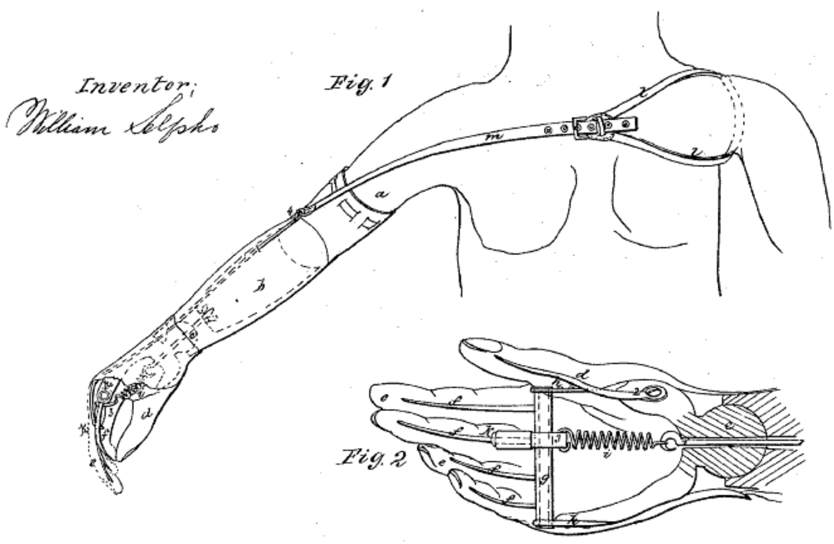
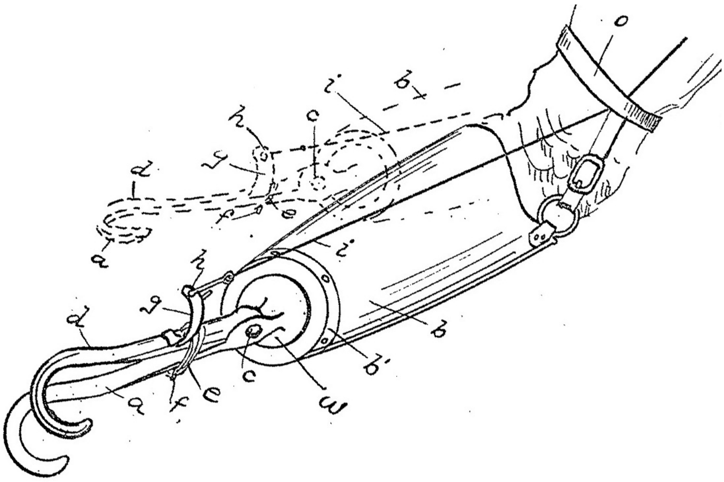


Figure 2.6: The first example of body-powered actuation; patented by Selpho (1857)PD.



(a) A drawing of Dorrance's first split hook (1912)[PD].

(b) A modern Hosmer Dorrance split hook. Image source: Hosmer Dorrance Corp. (2012). Used with permission.

Figure 2.7: Old and modern split hooks.

Even now, 74 years after the demonstration of the first myoelectrically controlled device (described in the next section), body-powered hooks and hands are still quite popular. The hooks have not changed much since 1919 (example in Fig. 2.7b), but more anthropomorphic body-powered prostheses have emerged (examples in Fig. 2.8). One reason for their popularity is that these devices are relatively cheap, simple and durable; important properties especially in developing countries and in countries with a sparsely distributed population and few prosthetic and orthotic workshops available to the users, as well as in countries without any public health service. Another reason is that they have sensory feedback, a concept often referred to as extended physiological proprioception (Simpson 1974). This allows for precise handling of small or fragile objects.

(a) (b)

Figure 2.8: Modern body-powered anthropomorphic prosthetic hand and harness. Image source: Otto Bock GmbH (2012).

2.2.2 Myoelectric Control

According to Childress (1985), the first known powered prosthesis was a German pneumatic hand, patented by Dahlheim (1915). Drawings of the first electric powered hand was published by Schlesinger (1919). Thirty years later Reiter demonstrated the first simple myoelectric prosthetic device (Reiter 1948), and other research groups published similar material shortly after (Berger et al. 1952; Battye et al. 1955; Bottomley et al. 1963). The focus of the prosthetics research was changed towards myoelectric control, and the first commercial myoelectric hands were available from the middle of the 1960's (Sherman 1964).

Myoelectric control is by definition the control of a prosthesis or other system through the use of "muscle electricity": The term *myo* comes from the greek word *mys* (muscle). The origin of the myoelectric signal; the "electrical activity produced by a contracting muscle", is well described in literature (Childress 1992; Lovely 2004b).

The electromyogram (EMG) may be picked up by pairs of internal wire- or needle electrodes, implanted electrodes or surface electrodes. Several types of surface electrodes exist; gel-type electrodes using an electrolyte interface to the skin in order to lower the electrical resistance, and the "dry" type made from metal (typically stainless steel). For long-term use, such as in prosthesis control, "dry" surface electrodes are the only practical solution available (Childress 1992).

One may look at the EMG signal as a measurement of the prosthesis user's intent, since the muscles are the actuators performing the user's intended movements (in able-bodied). Determining user intent from the electrical activity recorded from the brain is not yet solved, as illustrated by the large effort being directed to the goal of brain machine interfaces. On the other hand, when we measure on nerves or muscles, some of the interpretation has already been performed by the body of the subject, thereby simplifying the task. This is, however, still challenging: The surface EMG measurements are measuring on the outside of the limb and will thus contain a mixture of signals from the muscles nearby. Even when placing the electrode directly above a certain muscle, one may pick up signals from other muscles. The interaction between muscles is called *crosstalk* and needs to be handled by the prosthesis control system.

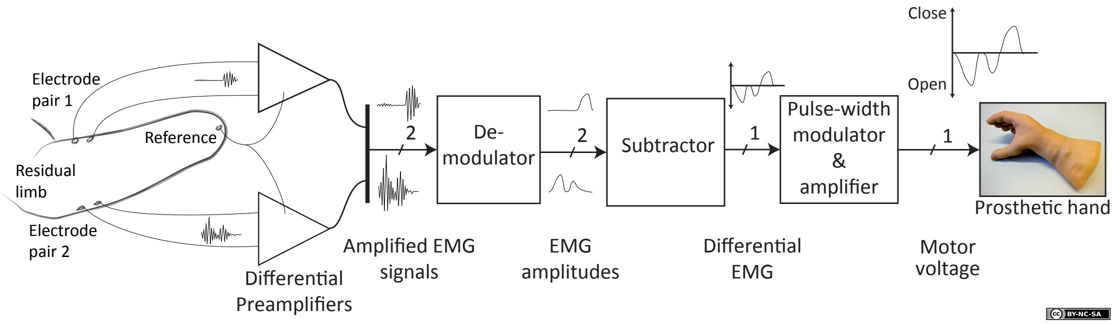


Figure 2.9: Simplified sketch of *traditional* proportional myoelectric control of single-function prostheses. The figure is licensed under a Creative Commons BY-NC-SA license (hereafter abbreviated to CC BY-NC-SA).

The electrodes may also pick up a large amount of noise from the environment. The prosthesis user is always capacitively coupled to the electromagnetic environment; the domestic line supply and electric devices. The amplitude of the resulting disturbances seen at the EMG electrode may even be larger than the EMG signal itself. This problem is common for all biomedical applications involving electrical measurements on the human body. Advanced differential amplifiers and signal processors have been developed in order to suppress these disturbances. An overview of the most common methods is offered by Lovely (2004a).

Fig. 2.9 shows a simplified sketch of *traditional* proportional myoelectric control (see Def. 2.3 in Section 2.3.3 and Paper F). This is very similar to the method suggested by Battye et al. (1955); Bottomley et al. (1963) and adopted by Horn (1963); Rothchild (1965); Alter (1966) in the early days of myoelectric control. It is now the *state of the art* in myoelectric control of *single-function* upper-limb prostheses.

Some prosthesis users are not able to achieve proper control of their device using the traditional proportional control: Sometimes the residual muscles are hidden behind scar tissue and do not offer good myoelectric signals. In those cases, the system can be simplified by using “On-Off” control (Reiter 1948) (see Def. 2.1). This can be made almost identical to traditional proportional control; by introducing thresholds in the demodulator. When there is only one proper electrode site available, the single-site system designed by Parker et al. (1977) can be used.

2.3 State of the Art in Control of Multifunction Prostheses

Since the end of the 1960's, it has been possible to use a powered elbow and a powered hand, or (later) a powered wrist and a powered hand, together in what we call dual-function or multifunction prostheses. During the last decade, several anthropomorphic multi-articulating hands have been introduced, two of them shown in Fig. 2.10. Both of these technological advances have generated a need for more sophisticated control systems than the traditional method presented in Fig. 2.9 which is only able to control a single function.

In order to describe the present and future systems for control of multifunction prostheses, it was determined that there was a need for a review of the terminology. A functionally partitioned model of the prosthesis control problem has been suggested (see Fig. 2.11) and is described in detail in Paper F, along with an unambiguous terminology. The layers of the model represent *principal functions* of the control system, i.e. not the physical software or hardware modules. The model is intended to fit all prosthesis control schemes, and thus three implementation examples are presented along with the model.

(a) iLimb Pulse, from Touch Bionics, Inc.

(b) BeBionic v1, from RSL Steeper.

Figure 2.10: Examples of commercially available multi-articulating prosthetic hands (CC BY-NC-SA).

Three important properties of myoelectric control systems are presented in Fig. 2.12: The Preprocessing layer, the Intent interpretation layer, and the Activation profile. In short, the Preprocessing layer is the collection of information from the user, a function typically implemented using sensors and signal processing. The Intent interpretation layer is the interpretation of user intent based on the available information from the Preprocessing layer. This is an essential part of a prosthesis control system, as this is the functionality defining the high-level control experienced by the prosthesis user. Finally, the Activation profile is one property of the Output layer of the system and distinguishes explicitly between proportional control and various on-off based schemes. These parts of a prosthesis control system are comprehensively described in Paper F, Part II.

In control of multifunction prostheses or prosthesis systems (such as a powered elbow used with a powered hand), sequential control is today the most common strategy. Typical implementations allow the user to scroll through a sequence of available states, by using co-contractions (Lovely 2004a) or a mechanical switch. Sequential control may be described as simple to use, in the sense that it allows the user focus on controlling only one motor function at the time. Nevertheless, prosthesis users commonly experience this control method as cumbersome and slow. This is not surprising, since a normal hand moves by using several muscles simultaneously (coordinated) among able-bodied.

In order to fully exploit the possibilities of these new multifunction prostheses, we need better and more intuitive control. The complexity and lack of robustness in today's control systems may be the reasons for why people still use the conventional hooks and body-powered prostheses.

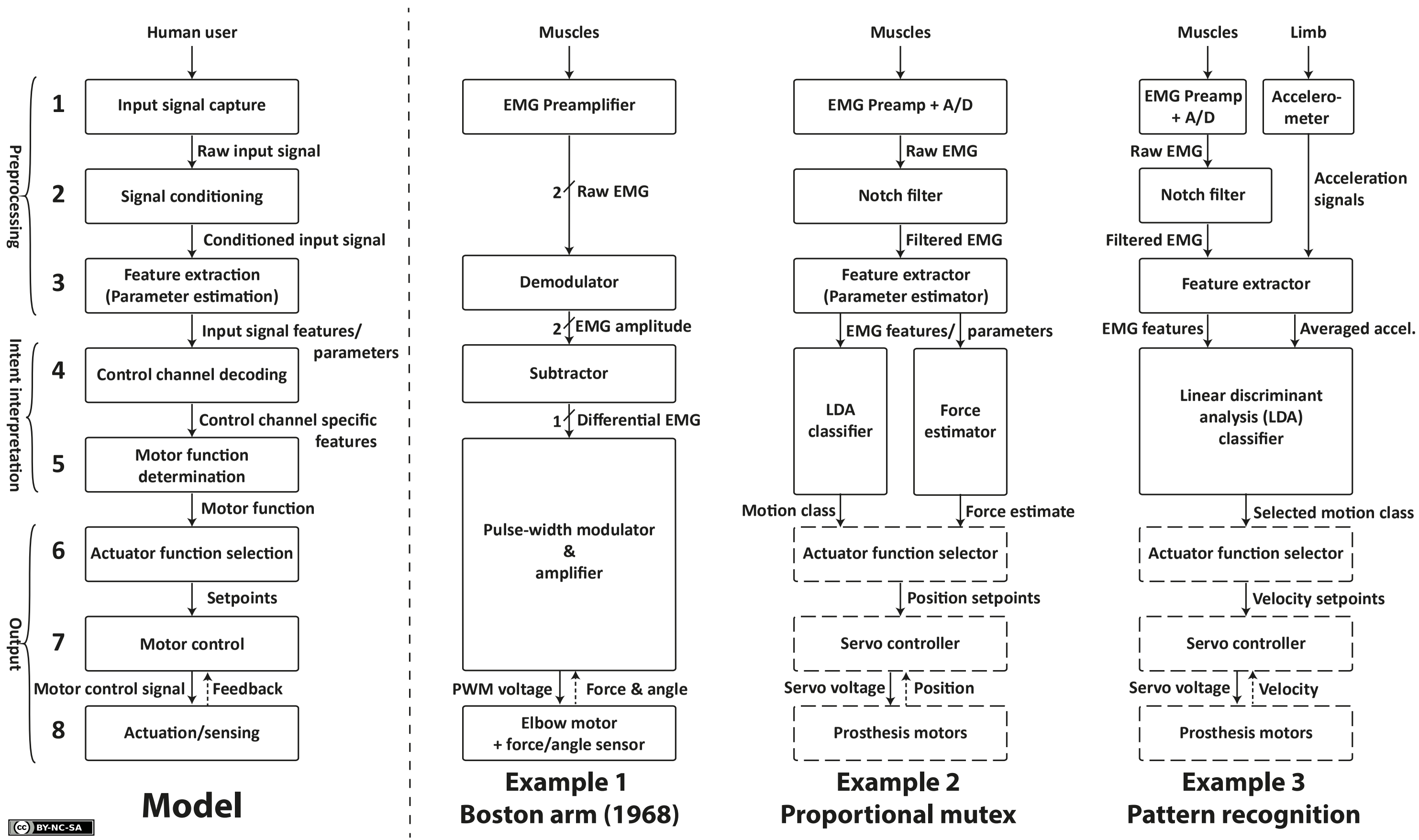


Figure 2.11: A functionally partitioned model and corresponding taxonomy for the prosthesis control problem. It is an extended version of the model proposed by Y. Losier (2009) and was published and described in Paper F. Three examples are given: Ex. 1 is the control system for the Boston Arm (Mann 1968). Ex. 2 is a proportional mutex control system, where levels 1–5 correspond to the research by Hudgins et al. (1993) and levels 6–8 (dashed lines) represent a possible implementation in a prosthesis. Ex. 3 is a multi-modal pattern recognition approach, where levels 1–5 are described by Paper C and the dashed lines represent a possible implementation of layers 6–8. Image source: Paper F (CC BY-NC-SA).

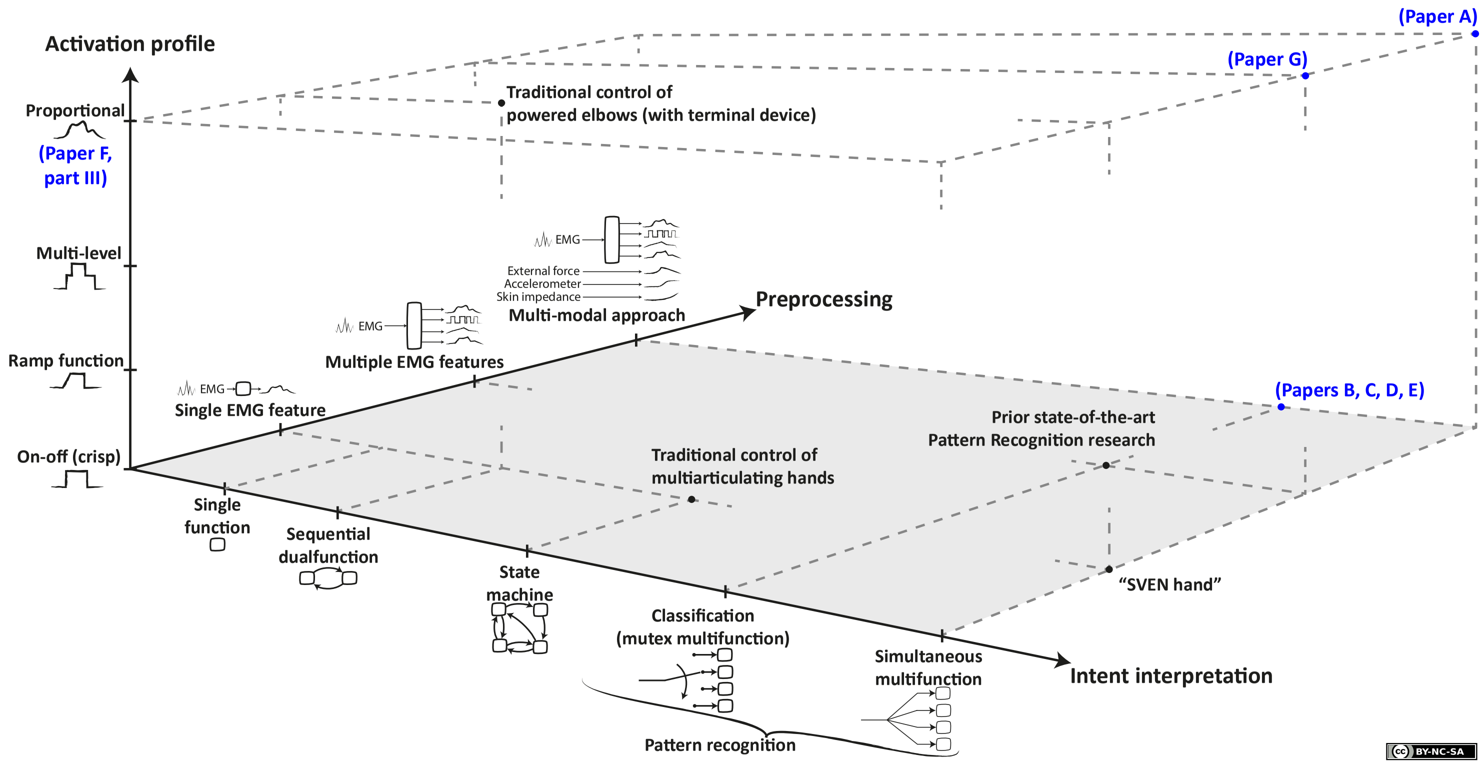


Figure 2.12: A 3-D representation of myoelectric control for upper limb prostheses. Indicated in the diagram are examples of traditional strategies in commercial dualfunction hands and arms and commercial multiarticulating hands, the "SVEN hand" (Almström 1977; Herberts et al. 1978; Almström et al. 1981), a typical modern pattern recognition system using crisp classification (Hudgins et al. 1993; Englehart et al. 2003), and the topics of Papers A–G. Part II of Paper F covers terminology for the whole domain, while Part III covers methods in the plane of Proportional control. The boxes on the *Intent interpretation* axis may either represent controlled motor functions (including the possibility of turning the function off) or classes (in the case of classification, where one of the classes is "prosthesis at rest". Based on an illustration from Paper F (CC BY-NC-SA).

2.3.1 On-Off Control

In Paper F we define *on-off control* as follows:

Definition 2.1. In the control mode called on-off control (also known as bang-bang control, crisp control or binary control), a function of the prosthesis is simply turned on or off (e.g. either constant speed in one direction, full stop, or constant speed in the other direction).

Comment 1: Previously this technique was inaccurately referred to as digital control, even if the control circuitry was mostly analog in nature. The term stems from the binary nature of the on-off control signals, which is a fundamental property of the signals in truly digital circuits. The use of the term "digital control" is discouraged in a prosthetics context to avoid confusion with modern digital control systems.

The SVEN hand, first demonstrated at Chalmers University of Technology (Sweden) in the 1970's, allowed the users to have simultaneous control of six motion classes: Hand open/close, wrist flexion/extension and wrist pro-/supination (Herberts et al. 1973; Almström 1977; Almström et al. 1981). It is one of the first known applications of pattern recognition in control of prostheses, along with the studies by Finley et al. (1967); Wirta et al. (1978) and Graupe et al. (1977). The SVEN system was based on a set of simple Bayesian perceptrons for myoelectric signals and controlled all motors simultaneously in an on-off fashion. The hand was not reliable nor portable enough for testing outside of the laboratories, but they experienced very promising results in clinical trials. Unfortunately the project was not continued, but researchers at Chalmers University still do research on upper limb prostheses (Ortiz-Catalan et al. 2012).

More recently, pattern recognition methods have become very common in research on control of multifunction upper limb prostheses, and the state of the art since 2003 is the control scheme suggested by Hudgins et al. (1993); Englehart et al. (2003). Their simple linear discriminant analysis (LDA) classifier and a set of four time-domain (TD) features from the myoelectric signal can offer high classification accuracies and appears to be relatively robust. Still, this system and a large amount of related studies on pattern recognition methods have not yet evolved into clinical use. Some of the challenges have recently been reviewed by Scheme et al. (2011b) and relate mostly to the differences between testing in a controlled environment (such as the laboratory) and real clinical use. This has previously been confirmed by Lock et al. (2005) (See Section 2.4 regarding assessment of prosthesis control systems).

When a prosthesis user interacts with the environment, there will be movements and forces acting at the interface between the electrodes and the skin, resulting in disturbances to the measured myoelectric signal. These effects are commonly referred to as motion artifacts or simply artifacts and are even more dominating in lower-limb prosthetics than in upper-limb. Researchers have known about them for a long time and developed techniques to reduce the artifacts by improving the electrode placement and stabilization, and by proper signal processing methods. The artifacts do, however, still represent challenges in myoelectric prosthesis control.

The work of this thesis includes development of methods for artifact cancellation to increase the system's robustness to variations during clinical use. This is reported in Paper A and in Papers B–E and discussed in Section 3.1.

2.3.2 System Training

Definition 2.2. *System training* is the training of a prosthesis control system to recognize input signals from the prosthesis user. This is often just referred to as *training*, *supervision* or *supervised learning* in pattern recognition.
Not to be confused with *User training*; training of the user's ability to control a prosthesis (see Paper F, Table I).

All pattern recognition methods need some kind of system training. Four different system training methods are described in Paper F. The most recent method is prosthesis guided training (PGT); the prosthesis is moving while the user follows the motions with the phantom limb. The strength of this method is that it is simple, quick and does not require an external computer. Thus the user can re-train the prosthesis whenever needed, for example by pushing a button and thereby start a training procedure. Prosthesis guided training was first demonstrated with on-off control by Lock et al. (2011); Simon et al. (2011b).

2.3.3 Proportional Control

In Paper F we define proportional control as follows:

Definition 2.3. *Proportional control* is exhibited by a prosthesis system if and only if the user can control at least one mechanical output quantity of the prosthesis (e.g. force, velocity, position or any function thereof) within a finite, useful, and essentially continuous interval by varying his/her control input within a corresponding continuous interval.

Comment 1: The term *essentially continuous* reflects the fact that most modern control systems are based on digital electronics, in which all continuous quantities are approximated by a finite number of increments. Usually, the small difference between adjacent quantization levels is imperceptible to the user; thus essentially continuous. A similar argument is valid for temporal discretisation whenever the sampling interval is sufficiently short to be neglible.

Comment 2: The notion of proportional control is not to be confused with a proportional controller as used within the control engineering field. In the latter case, a feedback controller generates a control signal proportional to an error signal within a closed loop, while in the prosthesis case, the term *proportional* relates to the system's forward path as such. To avoid ambiguity, we therefore discourage the use of *proportional controller* in a prosthetics context unless there is an explicit reference to a feedback controller. For the same reason, we suggest in general that the term *controller* is reserved for hardware or software modules that relate directly to actuator control, and rather use the more general term *control system* when discussing more high-level aspects of the problem.

Comment 3: Definition 1 does not require the relationship between control input and controller output to be strictly proportional in the mathematical sense, only that it must be essentially continuous. The rationale for this is that there is no objectively correct way to quantify a user's control input as a function of measured EMG signals or vice versa, and thus the mere notion of mathematical proportionality is irrelevant.

Comment 4: The term *useful* reflects that the functional relationship between user input and control system output must be of a suitable form. In particular, the effective amplification and the saturation limits of the system must be such that the user is in fact able to vary the output signal continuously in the entire output interval without the use of excessive muscle contraction or cognitive load.

Proportional control may be beneficial to the prosthesis user, for a number of reasons. Roesler (1974) claimed that proportional control is required for quick graping of objects, while at the same time having the possibility of slow and precise prehension. In a renowned work shortly after, Sörbye (1977, 1978) demonstrated that skilled users can successfully use an on–off system to lift and manipulate delicate objects, even while being blind-folded and deprived of acoustic feedback from the prosthesis. Three decades later, Lovely (2004a) claimed that the need to control the finger speed originally arose because of the slow motions in early prosthetic hands. Since the current prosthesis motors are much faster, speed control is not a critical issue any longer. For elbows, however, the range of motion is larger and the need for rapid, coarse, positioning is higher, while retaining the possibility of slow and fine control for accurate positioning of the terminal device. Thus, it was concluded that proportional control is useful for elbows but not critical for prosthetic hands. Alley et al. (2004), on the other hand, claimed that proportional control systems allow the wearer to vary the pinch force in a terminal device much more precisely than is possible with on-off control. The controversy around the necessity and appropriateness of proportional control in upper limb prostheses thus is still very much alive.

In a multi-function prosthesis control system, it is hypothesized that proportional control will enhance the user's control ability significantly, because the continuous relationship between muscular contractions and prosthesis response will allow for more rapid and high-fidelity corrections of movements that deviate from the user's motor intent.

A comprehensive literature review on proportional myoelectric control is presented in Paper F, and an updated chronological overview of the literature is shown in Fig. 2.13. As indicated in Fig. 2.13, the newest publications (on proportional control) have focused on simultaneous proportional control (Ameri et al. 2012; Hahne et al. 2012; Jiang et al. 2012b; Muceli et al. 2012; Pulliam et al. 2012).

A simple example of proportional myoelectric control is a system in which the EMG from flexors and extensors of the user's forearm is measured, amplified, filtered and smoothed by two active electrodes. This provides estimates of EMG amplitudes that can be sent to a hand controller. After applying thresholds to remove uncertainty at low contraction levels, the controller sets a voltage applied to the motor that is proportional to the contraction intensity (Sears et al. 1991). This functionality is essentially offered by several manufacturers of commercial prostheses. A simplified sketch of such a system for control of single-function devices was presented in Fig. 2.9.

For control of multi-function devices, however, the system needs a method to coordinate the motions. As illustrated on the Intent interpretation axis of Fig. 2.12, the there is a number of different approaches to this task.

Simultaneous control, as opposed to *sequential control*, *state machines* or *classification* (see Fig. 2.12 and Definitions in Paper F), is hypothesized to be the most intuitive control system to handle for the prosthesis user. Sequential control is on the other hand deemed

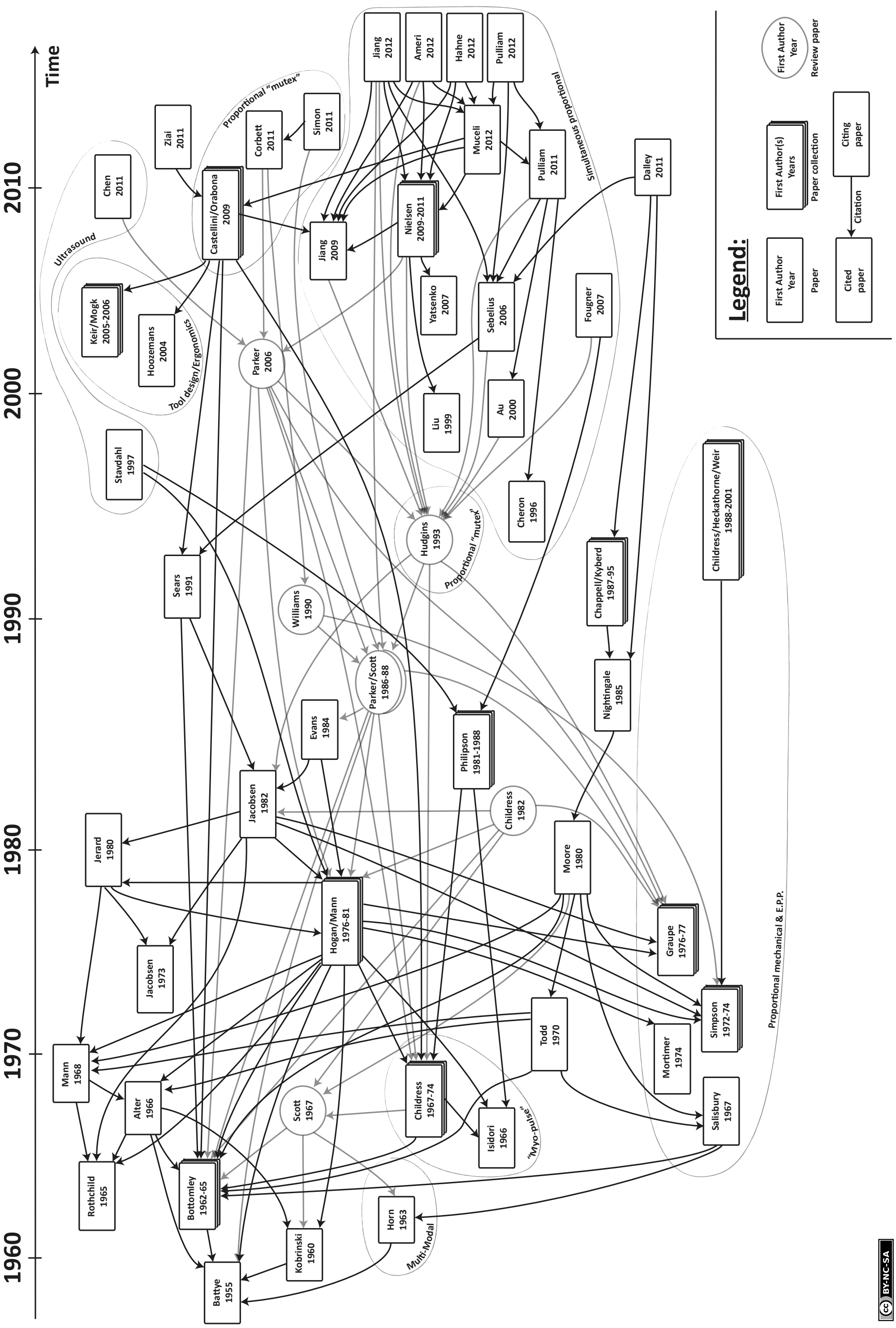


Figure 2.13: A chronological representation of literature on proportional myoelectric control for upper limb prostheses. Only the first author is mentioned in the figure. Papers that only refer to other papers on proportional control, usually in a review, without actually using proportional control, are shown as gray "nodes". Based on an illustration from Paper F (CC BY-NC-SA).

as slow and inconvenient by many users, but it is today the only method available in commercial multifunction prostheses. A real-time implementation of a control system with simultaneous proportional myoelectric control, for a dualfunction prosthesis, is presented in Paper G. The method includes prosthesis-guided training, and the assessment required development of a novel prosthesis socket equivalent for use by normally-limbed subjects.

The work presented in here is related to a number of recent publications on simultaneous proportional control (Ameri et al. 2012; Hahne et al. 2012; Jiang et al. 2012b; Muceli et al. 2012). The main differences are that Paper G contains practical testing of such a system implemented on a commercially available prosthesis, and the use of prosthesis guided training. Other related research includes lower limb exosceletons (Ferris et al. 2009), tool design and ergonomics (Hoozemans et al. 2005; Keir et al. 2005; Mogk et al. 2006) and kinesiology (Cheron et al. 1996; Liu et al. 1999; Au et al. 2000).

2.4 Assessment of Prostheses and Prosthesis Control Systems

Assessment and outcome measures for prostheses and prosthesis control systems is a challenging task. There is a large variety of methods being used to assess the outcomes in the clinic. In 2008–2009, an upper limb prosthetic outcome measures (ULPOM) group was established in order to create a systematic approach for the task and standardize the methods (Hill et al. 2009). The International Classification of Functioning, Disability and Health (ICF) model defined by World Health Organization (2002) was used to identify the tools that can be used to measure outcomes in various parts of the prosthetics research and profession.

According to the ULPOM group, the domains *Function*, *Activity* and *Participation* of the ICF model can be related to the situations *research*, *development*, *clinical assessment* and *daily use* (Hill et al. 2009). When assessing myoelectric control systems, the corresponding progression can be identified as going from a static, restricted laboratory setting with off-line classification or mapping, through real-time implementations and testing of prostheses in simple tasks, towards general dynamic movements in the clinic and daily tasks at home.

It is usually not practically possible for researchers to test their novel prosthesis control systems in the home of a prosthesis user. Still, it may be worth considering to move towards the Activity domain by performing the tests in a less restricted setting than sitting in front of a computer, and by implementing and testing the system on a prosthesis rather than a virtual environment inside the computer. Whenever the system is available for real-time testing with a prosthesis, it is anyhow still a challenge to assess the prosthesis.

For assessment of simultaneous versus sequential control, the ULPOM group concluded that: “It would be beneficial to researchers to be able to evaluate the use of

Figure 2.14: Equipment used for assessment in the function and activity domains. *Left:* The Original Rolyan Graded Pinch Exerciser with red clothespins. *Right:* The SHAP kit. Image source: Paper G (CC BY-NC-SA).

simultaneous control of various components versus seamless sequential control. This is hard to achieve at present." During the study presented in Paper G, an effort has been made in order to find relevant tools for such a comparison. These two assessment methods are used:

SHAP (Southampton Hand Assessment Procedure) is a clinically validated test. The kit is shown in Fig. 2.14 and the procedure consists of:

- **Manipulations on 12 abstract objects:** Six lightweight (from wood) and six heavy (from stainless steel) of the same shapes, each one requiring a different type of grip.
- **14 activities of daily living:** Picking up coins, undoing buttons, food cutting, page turning, opening a jar lid, two types of water pouring, lifting of three

different objects, and using a key, a zipper, a screwdriver and a door handle).

The kit is placed on a table horizontally aligned with the subject's hip, and body movements are not restricted. Each task is self timed and the functional score is based on the task completion time, relative to a normal population. The protocol is described by Light et al. (2002); Kyberd et al. (2009) and has its own website (Southampton Hand Assessment Procedure 2012).

The Clothespin Relocation Task has been commonly used by researchers in Chicago (Miller et al. 2008; O'Shaughnessy et al. 2008; Simon et al. 2012) and demonstrates a prosthesis system's ability to handle a task where at least two motor functions are needed. This test is therefore particularly useful in testing of multifunction control systems. Fig. 2.14 shows the clothespin kit. The protocol of the clothespin test is included in Paper G.

In assessment of a prosthesis control system, these tools will still have their scores affected by the user, the prosthesis and the prosthesis socket. Thus, it is necessary to keep the prosthesis and the socket design constant during a comparison study, and to have a significantly large number of users or test subjects involved. Different control systems can then be compared. When interpreting the results, one must still remember that the other variables affect the overall scores.

Chapter 3

Discussion

The thesis title is *Robust, Coordinated and Proportional Myoelectric Control of Upper-Limb Prostheses*. In Chapter 2, the reader was introduced to the last two terms: *Upper-Limb Prostheses* and *Myoelectric Control*. In this chapter, we will look at the definitions of the remaining terms; *robust*, *coordinated* and *proportional*, and discuss how these topics have been addressed through the work on artifact cancellation, simultaneous control and proportional control presented in Chapter 5. The validity of the results is also discussed.

3.1 Robustness and Artifact Cancellation

External disturbances to EMG signals are often referred to as *artifacts*.

Definition 3.1. *Artifacts* (in biomedical instrumentation) are those parts of a signal that originate from some source other than the one being studied.

An example of an artifact is demonstrated in Fig. 2 of Paper A. Another typical example is the unwanted interference of EMG from surrounding muscles when measuring ECG (electrocardiography). *Artifact cancellation* is the removal of such signal artifacts.

The related term *robust* is usually defined using words like “strong”, “sturdy” or “vigorous”. However, in decision making and software design, such as the intent interpretation layer of a prosthesis control system, robust may be defined more specifically as “capable of performing without failure under a wide range of conditions” (Merriam-Webster 2012).

Robustness in a prosthesis control scheme without feedback (other than visual) from the prosthesis to the prosthesis user is challenging to achieve. However, one common strategy for such a situation is to measure the conditions (e.g., a disturbance) affecting the system, and use this measurement to reduce the adverse effects (e.g., suppress the disturbance). In control engineering it is referred to as feedforward control, and when describing an Intent interpretation method (Fig. 2.11) it may be expressed as “gather as much relevant information as possible in order to make good decisions”.

As proposed in Section 2.3.1, multifunction prostheses may have degraded performance when the user is interacting with the environment, e.g. when the limb is moved to other positions than the one used in the training, and when the user is tired or sweaty. Thus, a natural direction of the research is to measure the interaction with the environment, the limb position, the fatigue and the sweat (through skin impedance measurements). During the work of this thesis, two of these problems have been addressed:

- In Paper A, force measurements have been added to the EMG electrodes in a “multi-modal unit” (MMU), in order to measure and cancel artifacts from external forces. The external forces may come from the mass of the prosthesis itself or from an object that the prosthesis is holding or touching.

- In Papers B–E, accelerometers were added to the forearm of the subjects in order to measure the limb position (by finding the direction of gravity) and thereby increase the system’s robustness against changes in limb position. The study was based on data from normally-limbed subjects, and the intent interpretation method used crisp classification (linear discriminant analysis).

Both of these lines of research are being continued, as studies of external forces affecting prosthesis control (Stavdahl et al. 2011, to be submitted), and as further studies of the limb position effect (Geng et al. 2012; Khushaba et al. 2012), also for prosthesis users (L. Chen et al. 2011) and proportional myoelectric control (Jiang et al. 2012b). There is a great potential in continuing this development, by adding more sensor modalities to the system, in order to cancel the artifacts in the EMG signal.

As proposed by Scheme et al. (2011a,b), the multi-modal approach is not the only way towards robustness of the prosthesis control system: Already when extending the system training protocol to include more variation (e.g. multiple limb positions, dynamic motions, external forces or fatigue), the system may become more robust towards those situations. This approach has also been taken by Hargrove et al. (2008). In order to increase a pattern recognition system’s robustness towards electrode shifts occurring during prosthesis use, it was suggested to extend the training protocol, by including a variety of possible shifts.

In summary, if one wants a prosthesis control system to perform well in a variety of situations, such as in the clinic and at home, one may need to include at least some realistic situations in the training protocol. For each situation added to the training protocol, the system training may become more time-consuming and exhaustive to the prosthesis user. The inconvenience may be reduced by letting the prosthesis user add more data to the training set, for example through the use of prosthesis guided training, whenever needed. That would minimize the need for long training session in the clinic. Every time the user adds training data, the control system may become more robust.

By the use of additional sensor modalities, a prosthesis control system may become even more reliable. Actually, as long as the system training method and the optimization criterion is well chosen, and the system is fast enough to handle the extra amount of data to process, the added measurement cannot degrade the reliability of the system. If a sensor (whether it's an EMG sensor or a different sensor modality) appears to be of no use during system training, the system itself will decide to ignore that measurement. The relative importance of each sensor is determined by the system training.

The work presented on artifact cancellation and robustness of the prosthesis control system is related to publications on sensor fusion and multi-modal approaches in other fields of study, such as speech recognition (Chan et al. 2002), activity monitoring (Roy et al. 2009), sign language recognition (Li et al. 2010) and person authentication (Duc et al. 1997).

3.2 Motor Coordination

The most recognized definition of the term *motor coordination* is the one by Bernstein (1996, p. 41):

Definition 3.2. *Coordination* is overcoming excessive degrees of freedom of our movement organs, that is, turning the movements organs into controllable systems.

In order to simplify tasks to the human mind, the body can coordinate the movement organs so that the person can relate to fewer degrees of freedom. This is similar to how the prosthesis manufacturers let the user choose a "grip pattern" (e.g., power grip, pinch grip or key grip) instead of controlling every finger joint one by one. The motion of each finger is coordinated with the motion of other fingers – their motions are functions of each other, with only one remaining variable to be controlled by the user (the choice of grip pattern). Thus, the coordination of these movements happens in the prosthesis - not in the neural system of the prosthesis user.

It is challenging to define what "good" motor coordination corresponds to, especially when discussing the coordination of movements in a prosthesis. One possibility is to instead use the definition of *dexterity*, again from Bernstein (ibid., p. 242):

Definition 3.3. *Dexterity* is the ability to solve a motor problem correctly, quickly, rationally, and resourcefully.

To activate a specific grip pattern in a multifunction prosthesis system is not necessarily a simple task. Usually, the prosthesis user needs to “scroll” through a series of available grip patterns in a sequential manner, using co-contractions detected by EMG (or a mechanical trigger). Similarly, in a system with a powered wrist and a powered hand, one usually needs to switch between hand and wrist control, i.e. another sequential control system. This method is commonly deemed as slow and inconvenient by prosthesis users.

Although the finger movements may be coordinated internally in the prosthesis (as described above), the overall motion of the hand does not satisfy the above definition of dexterity (or dexterous). It is neither well timed, smooth, nor efficient, compared to how able-bodied coordinate their motions. It may, however, be possible to let the prosthesis user take care of the motor coordination in their upper limb prosthesis, in a similar way - if the motor functions are allowed to be controlled simultaneously, and if this control system is reliable enough to let user’s body learn how to coordinate the available motor functions.

The feasibility of simultaneous control was demonstrated in the SVEN study in the 1970’s (Almström 1977; Herberts et al. 1978; Almström et al. 1981), and in Paper G. Whether it is possible to offer a simultaneous control strategy that is robust enough for clinical use, has not yet been proved.

3.3 Simultaneous Proportional Myoelectric Control

As a preparation to the experiment presented in Paper G, the literature on proportional myoelectric control was reviewed (cf. Paper F). It was discovered that few publications exist regarding the choice of system training method and the composition of the training data set – especially for simultaneous control. In this context, the notion of outcome measures is essential. By definition, system training involves optimization, and the quality of the results depends heavily on the choice of appropriate optimization criteria.

The training method used in Paper G is based on prosthesis guided training (Section 2.3.1). Use of this method has previously been reported only for crisp classification (Lock et al. 2011; Simon et al. 2011b), so further adaptations were needed in order to be suitable for simultaneous proportional control. The main reason was the subject’s need to sense, or observe, the forces or the speed of the movements demonstrated by the prosthesis: While these variables can be controlled in an on-off manner in training of a crisp classification scheme, they need to contain continuous movements in training

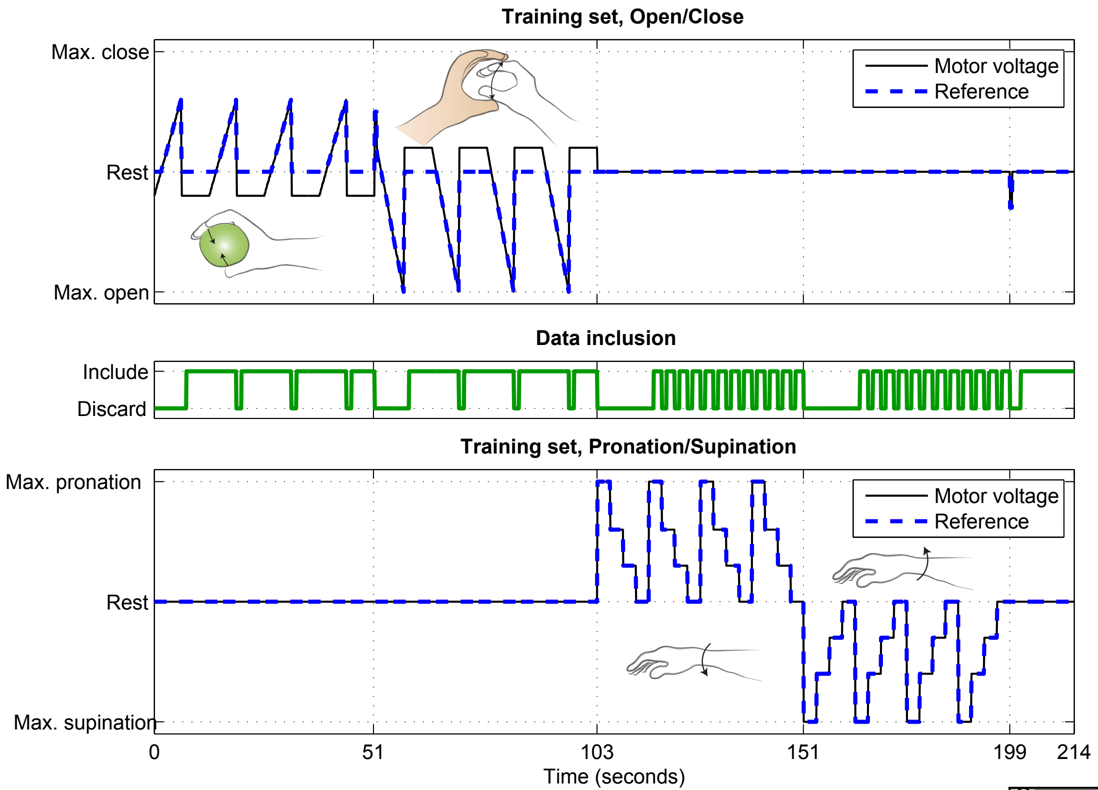


Figure 3.1: Prosthesis guided system training for simultaneous proportional control. The upper plot shows the voltage applied to the hand motor, and the lower plot shows the voltage applied to the wrist rotator. Some parts of the training procedure were discarded, as indicated by the boolean variable in the middle plot. The hand (colored) and prosthesis (white) sketches illustrate how each phase of the training was performed (more thorough explanations can be found in Paper G, Section II-F). Image source: Paper F (CC BY-NC-SA).

of proportional control. In order to follow those movements, the subject needs to perceive the motion. The suggested method is illustrated in Fig. 3.1. “Hand close” force was perceived by observing the prosthesis squeeze a soft rubber ball, and the subject tried to copy the force by using the finger flexors and/or wrist flexors. “Hand open” was trained in a similar way, by gripping around the prosthesis with the opposite hand in order to feel the applied force. Wrist rotations were trained by observing speed rather than force.

During the development of this training method, several movement patterns were considered. Initially, sinusoidal motor voltage patterns were applied, both for the hand motor and for the wrist rotator. It was found that the sinusoidal pattern was difficult to follow for the subjects. For hand open/close, the reason was mainly that the hand stopped when it arrived at its end-point (e.g., fully closed or fully opened), before the

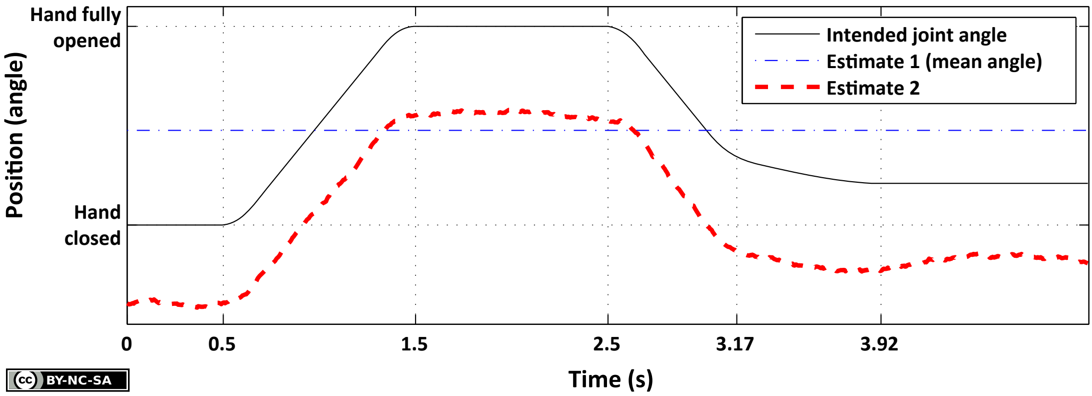


Figure 3.2: Illustration of an estimation problem when RMS error is used as performance measure. The data are generated as an example. Image source: Paper F (CC BY-NC-SA).

reference was at the boundary value (max. close or max. open). Without any feedback from the prosthesis, the system would not know whether the hand is moving or has stopped. In the end, a triangle-shaped voltage pulse was applied to the hand motors, and it was tuned so that the motors would stop approximately when the maximum voltage was applied. This is not optimal, but it works as a preliminary solution. For the wrist motors, similar triangle-shaped pulses were also tried as reference, but it was found difficult to know when the speed was at its maximum value. Thus, steps of three easily distinguishable levels (high, medium and low speed) were applied, the maximum value first. All motions were repeated four times. This allowed for the test subjects to "'practise" during the first motion, which was not recorded, in order to be well prepared for the next three repetitions. The training method lasted for approximately five minutes, including short breaks between each type of motion.

In Paper F, it was determined that a training set for simultaneous proportional control needs to contain simultaneous movements, unless some kind of interpolation is being used. The training method would then become more realistic – an important property of system training, according to previous findings in this thesis. During the development of this method, however, initial trials indicated that simultaneous motions in the prosthesis were too difficult to perceive for the test subjects. It may still be possible to use simultaneous movements in system training, but that topic needs to be addressed in future research.

The need for better training methods, revealed in Papers F–G, confirms the findings discussed previously: The training methods need to be improved in order to exploit the modern multifunction prostheses, for proportional control as well as for crisp classification and other control strategies.

The optimization criterion used to train the linear mapping function (in Paper G) was the minimization of root-mean square (RMS) error. In Paper F, it was demonstrated that RMS error is not necessarily a good measure of performance in a prosthesis control system; as illustrated in Fig. 3.2, where we have compared two possible joint angle estimates with an intended joint angle. These estimates can, for example, be based on signal features of measured EMG. Estimate 1 is the mean joint angle (dash-dotted), i.e. at a fixed joint angle, while estimate 2 (dashed) has approximately the correct shape but contains an offset. It is obvious that estimate 2 is more useful for a prosthesis actuator control signal than estimate 1, although estimate 2 has a larger RMS error: If used as a reference to the prosthesis motors, estimate 1 would not give any motion at all, while estimate 2 would give approximately the correct motion, at the wrong angle. With visual feedback a prosthesis user may be able to perceive and correct that error, in a conscious or unconscious manner.

For a non-linear estimator, e.g. a multilayer perceptron network with non-linear activation functions, the scenario illustrated in Fig. 3.2 is conceivable. It is, on the other hand, impossible when using a linear mapping function; A linear combination of a set of EMG signals or EMG signal features will never become a straight line (as estimate 2 in Fig. 3.2), as long as the linear mapping has non-zero elements.

The study presented in Paper G is of preliminary nature. The data collection for assessment of the four control strategies to be compared was a time-consuming process: The total recording time was approximately 20–40 hours during a period of 3–4 weeks for each subject. This is the reason for including only two normally-limbed subjects so far. Due to that number, conclusions cannot be drawn about the overall performance of this system. Even so, the results indicate that the three modern systems with simultaneous proportional control, mutex proportional control and crisp classification may all be superior to the conventional, sequential proportional control system in practical use. Future comparison studies with more subjects or prosthesis users are wanted, in order to confirm the findings of Paper G and separate the performance of the three modern systems.

During the Clothespin Relocation task with sequential proportional control, frequent use of *compensatory movements* (e.g., rotating the whole upper body instead of rotating the wrist) were observed. This indicates that compensatory movements may still be the fastest way to complete the test for this control system - even though the Clothespin Relocation task is designed to encourage the use of two motor functions. Thus, there might be a need for other test activities with a stronger dependence on using multiple motor functions, or ones with an explicit restriction of compensatory movements. On the other hand, we can not deduce from our results that all training effects (specifically the user's adaptation to the system) had died out by the completion of the last part of the experiment. It is possible that more extensive user training would increase the perceived functional performance to the point where the subject would instinctively prefer to utilize another prosthesis motor function rather than compensating with other body movements. Future research should address these issues.

The Clothespin Relocation task has previously been used by researchers in Chicago

(Miller et al. 2008; O'Shaughnessy et al. 2008; Simon et al. 2012). In Paper G it was chosen as one of the assessment tools, so as to compare the control strategies' ability to handle a task where at least two motor functions (e.g. hand open/close and wrist pro-/supination) are needed. During the experiments, it was observed that compensatory movements were frequently used, especially when using the sequential proportional control strategy. This indicates that compensatory movements may still be the fastest way to complete the Clothespin Relocation task for this control system - even though the test is designed to encourage the use of two motor functions. Thus, there might be a need for other assessment tools with a stronger dependence on using multiple motor functions, or ones with an explicit restriction of compensatory movements. However, one cannot deduce from the results that all training effects (specifically the user's adaptation to the system) had died out by the completion of the fifth session. It is possible that more extensive user training would increase the perceived functional performance to the point where the subject would instinctively prefer to utilize another prosthesis motor function rather than compensating with other body movements.

Both subjects in this study needed longer time (more than five sessions) to achieve stable scores in SHAP, while the results were relatively stable already after two sessions of the Clothespin Relocation task. For a continuation of this study, one should consider extending the number of recordings with SHAP. By increasing from five to ten sessions, one may achieve more consistent scores.

3.4 Terminology in Prosthesis Control Systems

The original purpose of the review of terminology, in Part II of Paper F, was to increase the readability of the other part of the paper (about proportional control): It was found impossible to describe and compare advanced control systems without first defining the various layers of the system with a model, suggesting an unambiguous terminology, and clarifying the relationships between different notions that are frequently confused in previous literature.

For example, it was revealed that the terms *degree of freedom*, *motion class* and *motor function* are commonly confused, although they are different. It was also found that the terms *intuitive*, *natural*, *dexterous*, *continuous*, *variable* or simply *myoelectric* control are all used to describe the method we have defined as proportional control, but they are sometimes used about other control strategies as well. This confusion may for example lead a research group to believe that they have invented something new, even though the problem has been solved and published previously, under a different name. Another example is when a prosthesis user and a clinician want to compare devices from different manufacturers, in order to choose the one that fits the need of the user. They might find it difficult to compare the devices, if the manufacturers use ambiguous terms. The confusion may be reduced if the whole community of researchers, clinicians and users of prostheses apply the suggested terminology.

One of the important results of the review is the model of the prosthesis control problem (Fig. 2.11), based on a simpler model proposed by Y. Losier (2009). The model was extended with the intention to fit all prosthesis control systems; also for lower limb. The proposed model can, for example, be used as a starting point for a modular control system, and the corresponding terminology can be used for communication between the different modules of such a system. Proper names for the variables sent between the modules, and documentation of the overall system by relating to the model, may increase the understanding of such a system, and it may simplify the developer's task of explaining the system for a clinician.

The proposed model is functionally partitioned, with eight layers representing principal functions of the control system. Thus, a physical implementation may miss one or more of these functions, and they may be applied in a different order. For example, a regular LDA classifier for on-off control will typically implement layer 4 (control channel decoding) and 5 (motor function determination) in a single intent interpretation module.

For simplicity, the model only contains the information flow from the human user to the actuators, except for the last step (feedback from the actuator to the motor controller). Use of feedback in powered upper limb prostheses has been reviewed by Childress (1980); Scott et al. (1980); Scott (1990). Targeted sensory reinnervation has shown promising results in recent studies (Kuiken et al. 2007; Marasco et al. 2009), inspired by the introduction of targeted muscle reinnervation (Kuiken et al. 2004; Kuiken et al. 2009). In order to describe a system with such feedback capabilities, the proposed model may need to be extended.

It remains to see whether the suggested terminology will be used. It has been emphasized that it is not the only possible choice, and that it is not necessarily complete. In the struggle to avoid confusing terms, difficult choices have been made, and it was attempted to include all the existing terms. In cases where expressions have been used in confusing ways, or in ways conflicting with other professional fields, new terms or redefined existing terms have been introduced.

In summary, the suggested terminology may stimulate the communication between researchers, clinicians, prosthesis users and other people involved in prosthetics. At the same time it may improve the understanding of the subject and stimulate to more structured research.

Chapter 4

Concluding Remarks

This chapter summarizes the work, defines relevant topics for future work and concludes the thesis.

Fig. 2.12 (p. 21) shows the relationship between the original publications in a three-dimensional figure. As previously discussed, the axes Preprocessing, Intent interpretation and Activation profile in the figure respectively correspond to the terms robust, coordinated and proportional from the title of the present thesis: Papers A–E have demonstrated how an increased robustness may be achieved, by using a multi-modal approach in the Preprocessing layer of the control scheme; Papers F–G have treated proportional myoelectric control, a property of the Activation profile; and Paper G has demonstrated simultaneous proportional control, which relates to the Intent Interpretation layer of the control scheme.

Other research groups focus on targeted muscle reinnervation (Kuiken et al. 2009), implantable electrodes (Merrill et al. 2011), osseointegration (Brånemark 1983; Jönsson et al. 2011) or feedback mechanisms (Kuiken et al. 2009; Marasco et al. 2009). The results of these activities are largely complementary to the material presented in this thesis.

Some of the contributions from this work may be applicable to myoelectric control of lower-limb prostheses. The work is also related to research on speech recognition, activity monitoring, sign language recognition and person authentication, lower limb exosceletons, tool design and ergonomics and kinesiology.

4.1 Conclusions

This thesis has addressed the topic of myoelectric control of upper limb prostheses, with particular emphasis on robustness, coordination and proportional control.

Through a complete review of the literature on proportional myoelectric control, the work of this thesis has united the historical contributions and offered a comprehensive overview of the topic for present and future researchers in the field. It revealed that methods for system training, both the choice of method and the composition of the training data set, need further research in order to achieve acceptable results with proportional myoelectric control.

Another important result is the discovery of an ambigous and incomplete terminology in literature describing prosthesis control systems. In order to address that, an unambiguous terminology was suggested. A model of the prosthesis control problem was also proposed. This contribution may have a positive impact on the communication between researchers and other people involved in prosthetics, and it may stimulate to more structured research.

Two versions of artifact cancellation were developed and demonstrated in this thesis. The suggested methods increase the prosthesis control system's robustness towards external forces and variations in limb position. The limb position effect was discovered, and also resolved, through the research presented. These contributions have triggered a development towards more robust and reliable control systems for prosthesis control, as demonstrated by several subsequent publications on related research.

The work of this thesis has demonstrated that simultaneous proportional myoelectric control of a dual-function upper limb prosthesis is attainable. The work involved development of a prosthesis socket equivalent for normally-limbed subjects which ensures near-isometric muscle contractions, as well as a novel use of prosthesis-guided training for proportional control. The results of the initial assessment trials are promising.

This thesis has contributed towards the long-term goal of offering an intuitive and robust control system to the end users of upper limb prostheses.

4.2 Future Work

Based on the work presented in this thesis, there are several possible future lines of research.

The study in Paper G may be extended by using an increased number of SHAP sessions and a larger number of subjects, preferably including some prosthesis users. It is tempting to incorporate a multi-modal unit (e.g. EMG-electrodes with accelerometers and force sensors) in such a study, in order to increase the robustness of the system. If this requires an extension of the system training protocol, for example by applying various amounts of external forces, an effort should be made to otherwise simplify the procedure.

It has been mentioned that the training method should contain simultaneous movements when used in a simultaneous control scheme. During the initial trials of the study reported in Paper G, it was found too difficult for the test subjects to follow simultaneous motions when they are being demonstrated by the prosthesis. However, this was a preliminary study. The training protocol needs further development, and one should not reject the possibility of having simultaneous movements in parts of the system training.

In Papers B–E, accelerometers were used as input to the control system along with the EMG signal features, and it was suggested to train the control system in a variety of positions. The intent interpretation method decides how to exploit the accelerometer data. A model-based approach may use the accelerometers in a more efficient way and simplify the training procedure. The accelerometers may be used directly to find the limb position (Fig. 3 in Paper B). By modelling how the limb position affects the myoelectric signals in the forearm, position invariant control can be provided with a reduced training set. Future research may exploit this possibility.

While the accelerometers were used to give information about the limb's orientation, they can also be used to measure the dynamic movements of the limb. By including such movements in the training protocol (Scheme et al. 2011a), one can exploit more information from the accelerometers, a solution that may increase the prosthesis control system's reliability during dynamic use. This may be important especially in simultaneous proportional control systems.

An effort has been made in order to standardize terminology in the field of prosthesis control systems. During that work, it became evident that also other parts of the prosthetics research and production may need a standardization of the terminology. For example, a proper definition of properties and grip patterns of multifunction hands could increase the communication between manufacturers, clinicians and end users.

A model of the prosthesis control problem was proposed (Fig. 2.11). This model covers only the information flow from the human user to the prosthesis actuators, except for the last step (feedback from the actuator to the motor controller). If continued research on targeted sensory reinnervation or other feeedback mechanisms is successful, as a feedback path from the prosthesis to the user, it may be useful to extend the proposed model in order to cover the complete system.

In Paper G, a need for better assessment tools was revealed. Specifically, tests with a stronger dependence on using multiple motor functions, or with an explicit restriction of compensatory movements, would be useful when comparing sequential and simultaneous control strategies. It would also be beneficial if the assessment included a measure of the mental burden on the subject. This is a challenging task, as indicated by Hill et al. (2009).

Chapter 5

Original Publications

This chapter contains six published papers in facsimile, as well as one submitted journal paper manuscript.

Paper A Cancellation of Force Induced Artifacts in Surface EMG Using FSR Measurements.

Published in the *Conference Proceedings of the Myoelectric Controls Symposium* (MEC 2008) at the University of New Brunswick, Fredericton, Canada, August 2008.

From “MEC '08 Measuring Success in Upper Limb Prosthetics,” Proceedings of the 2008 MyoElectric Controls/Powered Prosthetics Symposium, held in Fredericton, New Brunswick, Canada, August 13–15, 2008.

CANCELLATION OF FORCE INDUCED ARTIFACTS IN SURFACE EMG USING FSR MEASUREMENTS

Anders Fougner[1], Marthe Sæther[1], Øyvind Stavdahl[1], Peter J. Kyberd[2] and Jeremy Blum[2]

[1]Department of Engineering Cybernetics, Norwegian University of Science and Technology, Trondheim, Norway
[2]Institute of Biomedical Engineering, University of New Brunswick, Fredericton NB, Canada

INTRODUCTION

As multifunction prostheses become increasingly common, there is a need for improved control signal quality in order to control all the functions. Most signals commonly used for prosthesis control are sensitive to sweat, motion and external forces [1], which impairs prosthesis control performance.

We have developed a prototype surface electromyogram (sEMG) sensor with three built-in force sensing resistors (FSRs) for measuring the external forces, which may be used to cancel artifacts caused by these forces. The performance of the sensor as an estimator of muscle force is presented in this paper. The sEMG and FSR signals have also been tested individually, as a reference for the performance using the combination of these signals.

MATERIALS AND METHODS

The sEMG sensor unit was built from the metal electrodes of an Otto Bock 13E125 device, mounted with the original spacing and wired to an external preamplifier.

FSRs were chosen for force sensors due to their flatness and simplicity of use. Three individual FSRs allow both magnitude and position/direction of an external force to be estimated, factors both of which may be relevant for the artifact identification. Initial tests used an FSR component that was readily available. It is anticipated that with more appropriately sized sensors, the entire device will fit into a prosthesis socket. The sensors were sandwiched between two layers of acrylic glass using soft double sided tape (Fig. 1). The electrodes were attached to this structure with the reference electrode at the centre of the FSR array.

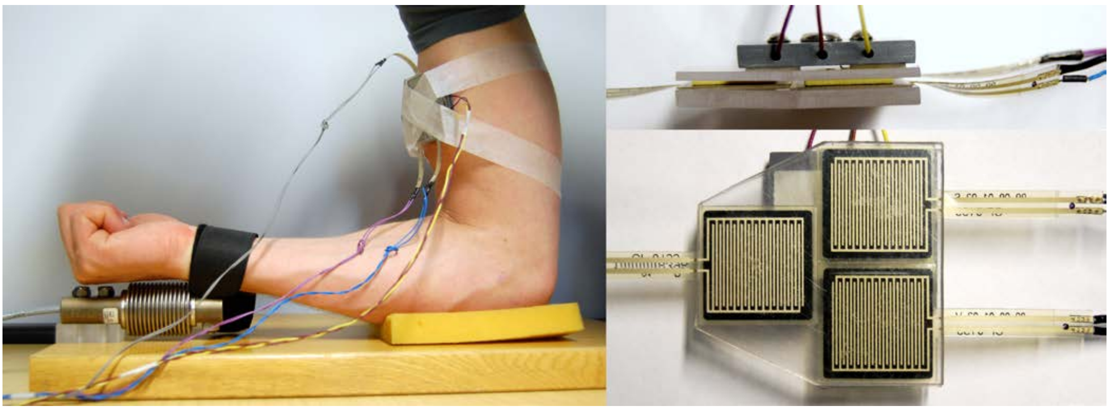


Fig. 1: Experiment setup and a close view of the sensor unit.

The device was taped to the *m. biceps brachii* of a healthy subject and tested by simultaneously measuring sEMG and FSR outputs while muscle contraction force was measured using a load cell (Fig. 1). The sEMG signal was pre-processed with a non-linear myoprocessor described in [2]. External forces in random directions were applied to the sensor during the measurements in order to induce artifacts. Data was collected at 218 Hz for approx. 50 s. Three data sets were acquired; a *training set* and a *validation set* collected immediately after each other, and a *test set* acquired after having removed and then reapplied the device to the subject’s arm.

From "MEC '08 Measuring Success in Upper Limb Prosthetics," Proceedings of the 2008 MyoElectric Controls/Powered Prosthetics Symposium, held in Fredericton, New Brunswick, Canada, August 13–15, 2008.

Multilayer perceptron (MLP) networks with different numbers of hidden nodes (2-25 nodes, 10 MLP networks of each size) were employed to estimate the muscle force based on sEMG and FSR signals. Following MLP training and validation, the best 50% of the MLP networks of each size were chosen for final assessment using the test set. A linear and a quadratic mapping function were also fitted to the training set for comparison.

RESULTS

Fig. 2 presents an example data set with all recorded data. Note the two central peaks in the FSR signals, which are not accompanied by peaks in the load cell signal; these represent artifacts. The result of the force estimation, using the test set and an MLP network and a linear mapping function, respectively, is presented in Fig. 3.

Fig. 4 shows the estimated against measured force for the test set after training and validating the MLP network. Note the presense of hysteresis in the FSR based estimate and the apparent treshold levels in the sEMG based estimates. Also note the presence of force artifacts in both sEMG based graphs, evident as significant force estimate values at approximate zero load cell force.

The root mean sqare error (RMSE) rates for the different combinations of sEMG and FSR as inputs are presented in Fig. 5. No reduction in RMSE was detected when increasing the number of hidden MLP nodes beyond n=4.

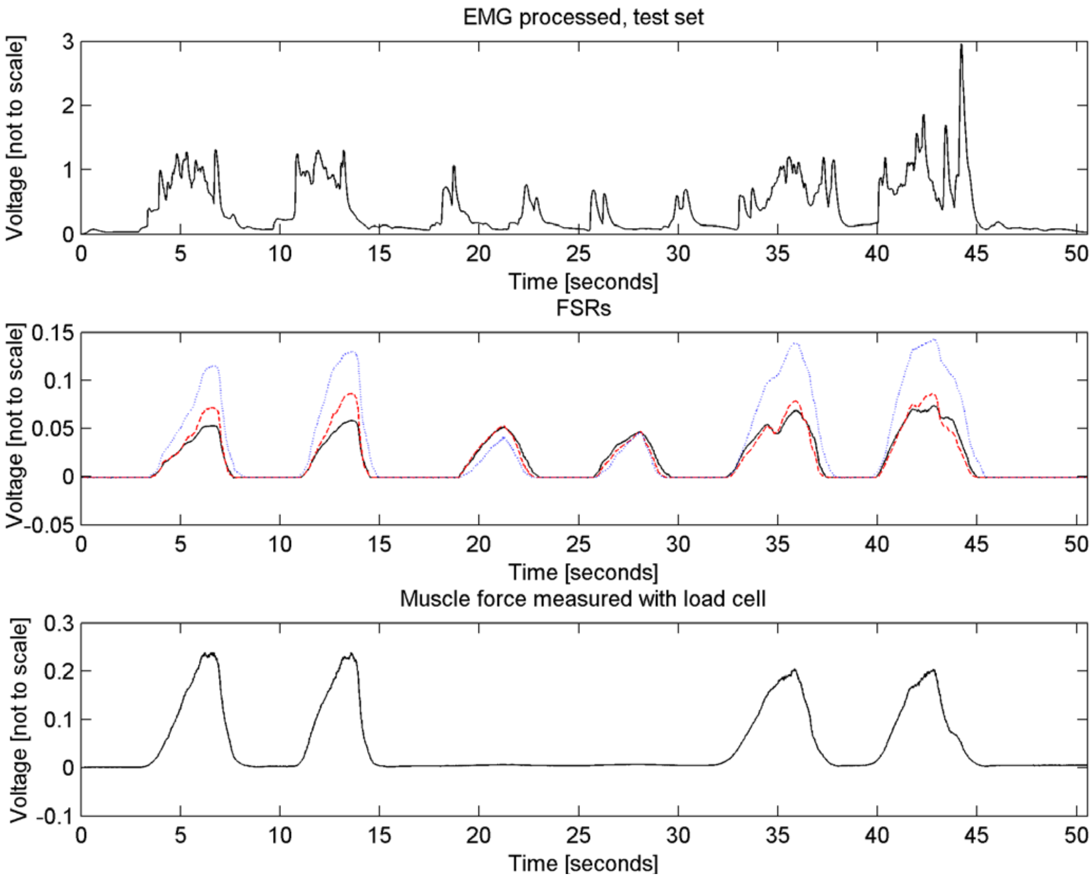


Fig. 2: Test set containing measurements from sEMG, 3 FSRs and the load cell.

From "MEC '08 Measuring Success in Upper Limb Prosthetics," Proceedings of the 2008 MyoElectric Controls/Powered Prosthetics Symposium, held in Fredericton, New Brunswick, Canada, August 13–15, 2008.

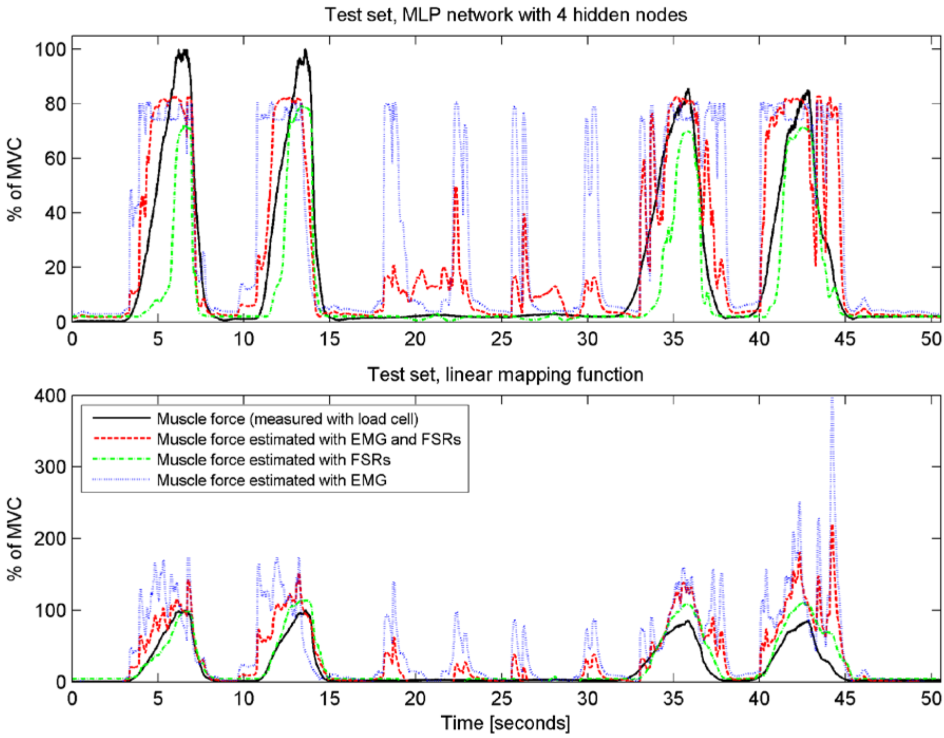


Fig. 3: Estimation results for three different test set inputs. Estimation using an MLP with 4 hidden nodes and a linear mapping function. Note different vertical scales; unit is percent of maximum voluntary contraction (MVC).

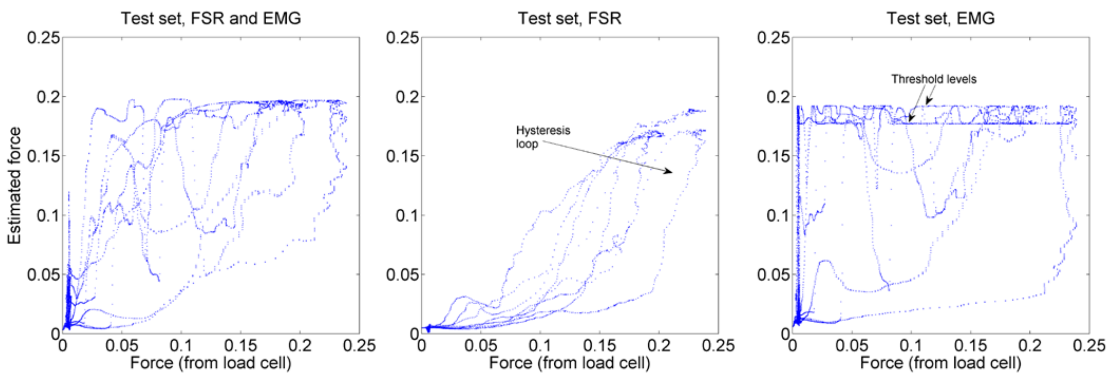


Fig. 4: Measured vs estimated force using an MLP with 4 hidden nodes. Same data set as in Fig. 3.

DISCUSSION

The results indicate that four hidden MLP nodes is a sufficient number to discriminate forces, as no improvement can be seen when increasing the MLP size beyond this point. The optimal MLP performed better than a linear estimator except when basing the estimate on FSR measurements alone, in which case the two techniques were equally successful.

From "MEC '08 Measuring Success in Upper Limb Prosthetics," Proceedings of the 2008 MyoElectric Controls/Powered Prosthetics Symposium, held in Fredericton, New Brunswick, Canada, August 13–15, 2008.

The quadratic estimator was fit to the training set without any validation. The results indicate that this has caused "overtraining" with respect to the training data, as evident from the estimator's inferior performance when subjected to the test set (cf. the caption of Fig. 5).

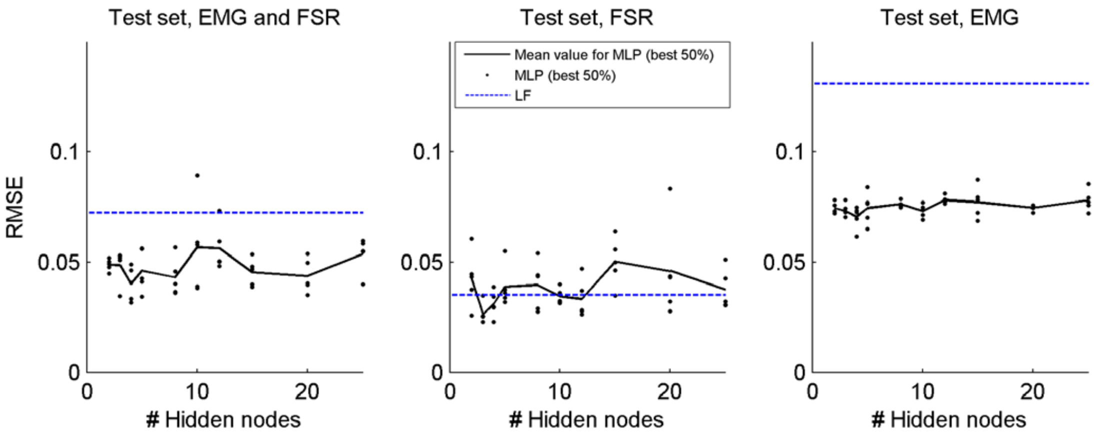


Fig. 5: RMS error rates for the same data set as in Fig. 3-Fig. 4. Corresponding values for the quadratic mapping function are 0.269 (EMG and FSR), 0.164 (FSR) and 0.152 (EMG).

It is noted that in Fig. 3 and Fig. 4, the FSR based estimates exhibit little or no artifact from external forces, which is at first a little surprising. In Fig. 2, however, it can be seen that the pure disturbance (i.e. the middle two "peaks") cause an equal response in all three FSRs, while when the muscle actually contracts, the FSRs yield different signal levels. Consequently, the estimator is able to distinguish these two signal sources.

In the upper graph of Fig. 2, the processed sEMG exhibits a transient response to the disturbance. This suggests that an optimal contraction force estimator should have a dynamic aspect rather than a purely static mapping property like the ones investigated in this study.

The results presented here are of a preliminary nature, and future study will assess the techniques using prosthetic sockets, real users and different myoprocessors. For example, the performance of a multi-FSR array inside a socket must be investigated, as the contact forces in that case may be different from those of the taped-on setup used in this study.

CONCLUSION

Measurements of contact forces exhibit promising properties for reducing force induced artifacts in conjunction with prosthesis control. The relative importance of sEMG and force measurements remain uncertain, and should be addressed in future work.

ACKNOWLEDGEMENTS

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Paper B Examining the Adverse Effects of Limb Position on Pattern Recognition Based Myoelectric Control.

Published in the *Conference Proceedings of the IEEE Engineering in Medicine and Biology Society's World Congress* (EMBC 2010) in Buenos Aires, Argentina, September 2010.

Paper C Resolving the Limb Position Effect in Myoelectric Pattern Recognition

Published in the *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, December 2011.

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Paper D A Multi-Modal Approach for Hand Motion Classification Using Surface EMG and Accelerometers

Published in the *Conference Proceedings of the IEEE Engineering in Medicine and Biology Society's World Congress* (EMBC 2011) in Boston, US, September 2011.

Is not included due to copyright

Paper E Resolving the Limb Position Effect

Published in the *Conference Proceedings of the Myoelectric Controls Symposium* (MEC 2011) at the University of New Brunswick, Fredericton, Canada, August 2011.

From "MEC 11 Raising the Standard," Proceedings of the 2011 MyoElectric Controls/Powered Prosthetics Symposium Fredericton, New Brunswick, Canada: August 14-19, 2011. Copyright University of New Brunswick.

RESOLVING THE LIMB POSITION EFFECT

Anders Fougner[1], Erik Scheme[2], Adrian D. C. Chan[3], Kevin Englehart[2], and Øyvind Stavdahl[1]

[1]*Department of Engineering Cybernetics, Norwegian University of Science and Technology, Trondheim, Norway*
[2]*Institute of Biomedical Engineering, University of New Brunswick, Fredericton NB, Canada*
[3]*Department of Systems & Computer Engineering, Carleton University, Ottawa ON, Canada*

INTRODUCTION

Electromyography (EMG) has been used as a control input for powered upper limb prostheses for decades. Alternative biosensors, like myokinemetric sensors [1], [2], mechanomyographic sensors [3] and accelerometers [4] have been used for upper limb pattern recognition in more general terms but have not produced accuracies acceptable for prosthetic use.

The desire to use a larger number of myoelectrode sites to facilitate control of multiple degrees of freedom has been counteracted by the added complexity, cost, space, and weight associated with additional sites. Thus, commercial upper limb prostheses today usually have only two electrode sites, while researchers continue to experiment with multiple sites [5]. An alternative to the uni-modal EMG approach for increasing the degrees of freedom is a multi-modal approach. Instead of adding additional EMG channels, it is possible to combine EMG and other sensor modalities (e.g., force sensors [6] or accelerometers [7]) in order to improve pattern recognition performance. Other examples of multi-modal solutions exist [8], [9].

In our previous work [10] it was shown that variations in limb position associated with normal use can have a substantial impact on the robustness of myoelectric pattern recognition. We proposed to solve this problem, hereafter referred to as the *limb position effect*, by training the classifier in multiple positions and by measuring the limb position with accelerometers. Applying these methods to data from normally limbed subjects, the classification errors were reduced substantially.

In the present study, we have examined the generalizability of the training set as a function of the number of training positions in the set. This makes it possible to define a minimum training procedure, in order to reduce the training time for the end user.

Finally, we have investigated accelerometers as a supplementary modality for EMG. Accelerometers are relatively cheap, small, robust to noise and easy to integrate in a prosthetic socket. This work examines the efficacy of accelerometers in comparison to adding expensive and space-consuming electrode sites.

METHODS

All experiments were approved by the University of New Brunswick's Research Ethics Board.

A. Population and Data Acquisition

EMG data corresponding to eight classes of motion were collected from 17 healthy normally limbed subjects (10 male, 7 female) within the age range 18 to 34 years.

Subjects were fitted with a cuff made of thermo formable gel (taken from a 6mm Alpha liner by Ohio Willow Wood) that was embedded with eight equally spaced pairs of stainless steel dome electrodes (EL12 by Liberating Technologies, Inc.). The cuff was placed around the dominant forearm (13 right, 4 left), proximal to the elbow, at the position with largest muscle bulk. A reference electrode (RedDot by 3M) was placed over the back of the hand. Two analog 3-axis accelerometers (Freescale MMA7260QT MEMS) were used to estimate limb position. The first accelerometer was affixed adjacent to the cuff on the forearm, over the brachioradialis muscle. The second was placed over the biceps brachii, aligned with the forearm accelerometer when the subject was reaching forward (see position P2 in Fig. 1). Both accelerometers were configured to have a sensitivity of 800 mV/g at a range of ±1.5 g, where g represents acceleration due to gravity.

The eight channels of EMG were differentially amplified using remote AC electrode-amplifiers (BE328 by Liberating Technologies, Inc.), and low pass filtered at 500Hz with a 5th order Butterworth filter. Finally, the six accelerometer channels and eight EMG channels were acquired using a 16-bit analog-to-digital converter (USB1616FS by Measurement Computing) sampling at 1 kHz.

Subjects were prompted to elicit contractions corresponding to the eight classes of motion shown in Table 1. Performance was evaluated using all eight classes, as well as a reduced set of five classes. This five class system only included classes C3, C4, C5, C6, and C8, which are representative of contemporary powered prostheses. The five class system is referred to as the *contemporary* system and the eight class system as the *advanced* system.

Table 1: Motion classes

C1.	**Wrist flexion**	**C5.**	**Open hand**
C2.	**Wrist extension**	**C6.**	**Power grip**
C3.	**Pronation**	**C7.**	**Pinch grip**
C4.	**Supination**	**C8.**	**Hand at rest**

Each contraction was sustained for three seconds and a three second rest was given between subsequent contractions. Ten trials were recorded in each of the following limb positions (P1–P5; as illustrated in Fig. 1), resulting in a total data set of [n subjects × 10 trials × 5 positions × 8 classes × 3 seconds], where n is explained in Section C.

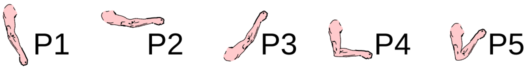

Fig. 1: Limb positions.

Subjects were instructed to perform contractions at a moderate and repeatable force level and given rest periods between trials to avoid fatigue. The average duration of the experiment (with 50 trials lasting 48 seconds each) was approximately 80 minutes per subject. Some patients noted minor shoulder (deltoid) fatigue.

B. Data processing

As this work represents an introductory examination of multi-modal pattern recognition, it was appropriate to test the effects using a known control scheme. Englehart and Hudgins [11] showed that simple time-domain (TD) feature extraction combined with a linear discriminant analysis (LDA) classifier can be used as an effective real-time control scheme for myoelectric control. Because of its relative ease of implementation and high performance, this system has been widely accepted and was therefore adopted in the present study. EMG data were digitally notch filtered at 60 Hz using a 3rd order Butterworth filter in order to attenuate any power line interference. Data were segmented for feature extraction using 250 ms windows, with processing increments of 50 ms. The TD features (mean absolute value, zero crossings, number of turns and waveform length) were extracted from the EMG data. Please refer to [11] for details of the feature extraction and the classification.

For each processing window, the average value of the accelerometer data was calculated. Where applicable, this feature (hereafter called ACCEL) was input to the LDA classifier separately or as an extension of the original feature set.

C. Data exclusion

Some of the subjects were not able to perform consistently throughout the data set. Similar phenomena occur in real-life situations where some individuals have great difficulty producing distinct EMG signals [12]. To ensure consistent data, subjects whose intra-position classification error exceeded 10% (five of the 17 subjects) were excluded from the study. This does not detract from the focus of this work; to ascertain the effects of position on performance. It simply eliminates possible confounding factors that may have been present with those subjects that did not perform well.

In two of the remaining 12 subjects, hardware problems caused erroneous accelerometer readings. Thus, 10 subjects were used in this study.

D. Classification

The following classifier training schemes were explored:

1) *Training in a single limb position*
TD features recorded from a single limb position were used to train the classifier. The classifiers were trained using data from the first five trials and tested using data from the last five trials.

2) *Training in multiple limb positions*
TD features recorded in multiple limb positions were concatenated and used to train the classifier. The classifiers were trained using a data set of reduced size per position, so that the total training set size was the same as in 1), in order to make the results comparable.

3) *Training with TD and ACCEL features*
TD and ACCEL features recorded in multiple positions were concatenated and used for motion classification. The data set was reduced in the same way as in 2) in order to make the results comparable.

E. "Leave-One-Out" training strategy

In order to investigate the generalizability of the training set as a function of the number of training positions in the set, the following procedure was employed. For each test position, all possible subsets of the remaining positions were applied as a training set.

F. Input selection

A signal feature selection scheme was chosen in order to examine which electrode sites and accelerometer signals would be most useful for the pattern recognition. Starting with just one sensor, the best one was chosen (based on the classification error averaged over all subjects and motion classes). It was then tested in combination with each of the remaining sensors, and the best combination was chosen before adding the next sensor. In this manner the sensors were added to the system one by one.

RESULTS

A. Training in a single limb position

Five different position-specific classifiers were trained; each one using data from only one of the limb positions, but tested using data from all positions. The resulting intra-position and inter-position errors are shown in Table 2.

Table 2: Intra- and inter-position classification errors for the advanced system, trained in a single limb position, and averaged across all subjects and classes.

Intra-position classification error	**3.8%**
Inter-position classification error	**21.1%**
Overall classification error	**17.6%**

B. Training in multiple positions

In Fig. 2, we present a comparison of how training in multiple positions affects the classification, for the *advanced* system. We have used the *Leave-One-Out* strategy as described in the *Methods* section, part E, in order to investigate the generalizability of the training set as a function of the number of training positions in the set.

Notice that the classification error improvement when increasing the number of training positions from one to two is larger than when increasing to three or four training positions.

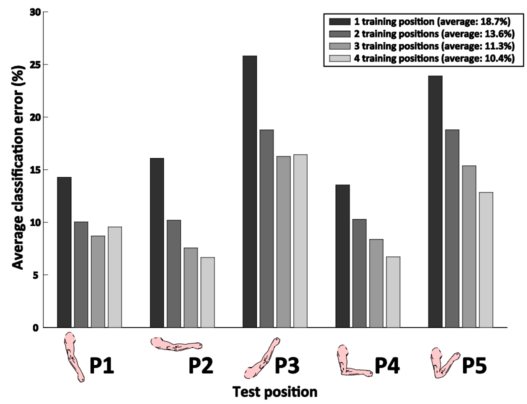

Fig. 2: Comparison of classification errors when testing in one limb position and training in all possible subsets of the remaining positions (the "Leave-one-out" strategy, as described in the Methods section, part E). Note that the training sets have been scaled so that they have identical size every time; independently of the number of training positions, by using subsets of the ten trials.

C. Relative importance of position information and surface EMG

The results of the input selection described in the *Methods* section, part F, are presented in Fig. 3. It is noteworthy that when adding new sensors one by one, the forearm accelerometer provides more novel classification information than even a second or third EMG electrode. It is also worth noting that the upper arm accelerometer is one of the least useful sensors. This is a desirable result as it would be difficult to justify including a sensor external to the forearm socket, and across the elbow joint.

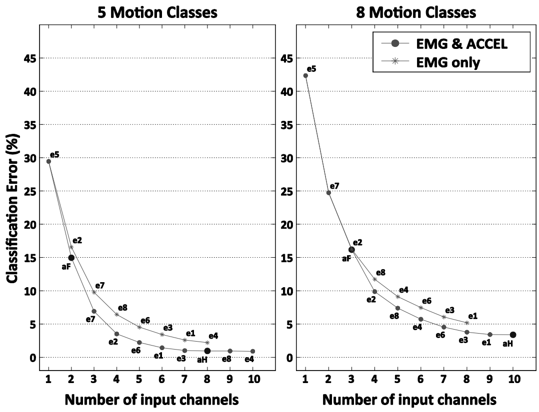

Fig. 3: Classification error as a function of selected input channels, for pattern recognition systems with 5 and 8 motion classes, choosing input channels among 8 electrode pairs (e1–e8) and 2 accelerometers (aF–Forearm, aH–Humerus).

For the *contemporary* system, the improvement flattens out after 4-5 electrodes and one forearm accelerometer (reaching an average accuracy of 98-99%). The advanced system can exploit 6-7 electrodes and one forearm accelerometer (reaching an average accuracy of 95-96%).

DISCUSSION

EMG TD features and training an LDA classifier in in a single limb position yielded an average intra-position error (3.8%) significantly lower than the corresponding inter-position errors (21.1%). These results indicate that EMG classification error is strongly dependent on limb position.

We have shown that the limb position effect can be partially solved by training the classifier in multiple positions. Since training in multiple positions can be cumbersome for the end user, it is however desirable to reduce the number of training positions. Therefore it is an advantage that most of the improvement is achieved already when increasing from one to two training positions (reducing the average error from 18.7% to 13.6%).

Previously we have also shown [10] that that it is important to have a training set containing a variation of elbow angle.

The accelerometer lends itself to being used in human-machine interfaces due to its small size, low cost, and simple mechanical and electrical interfaces. The absence of many of the disturbances often encountered in EMG sensors and similar devices makes it interesting as a supplementary sensor in hand motion classification systems, including upper limb prostheses.

The accelerometer does not provide an estimate of muscle force, but we have shown that it provides useful information that can supplement EMG signals. If one wants to improve a system originally having two EMG electrodes, a multi-modal approach can be taken. The results demonstrate that it is more advantageous to add an accelerometer affixed to the forearm (multi-modal approach) rather than increase the number of EMG channels (uni-modal approach).

Even though the limb position effect was discovered and observed in users in the clinic [7],[10], and was resolved for the normally limbed subjects in our study, it needs to be examined specifically for the end users. Gravitational and biomechanical effects of limb position will be different for prosthetic users compared to the normally limbed subjects of this study. As such, we are planning to extend this study to include prosthesis users.

This work is part of a larger investigation aimed at improving the practical robustness of myoelectric control. The present results indicate that facilitating position invariant myoelectric control through methods such as feature selection, data projection, multi-sensor systems, or by other means could be an important part of this larger work.

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Paper F Control of Upper Limb Prostheses: Terminology and Proportional Myoelectric Control—A Review

Published in the *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, September 2012.

Errata:

- On p. 667, Fig. 2; the indicator:

  Typical commercial elbow

  ...should have been:

  Typical commercial powered elbow with terminal device

  An improved figure is presented in Fig. 2.12 of this thesis.

- On p. 668, Section 2.D.4:

  "Simultaneous on-off control of six *motor functions* was first demonstrated in the SVEN hand in the 1970's."

  ...should have been:

  "Simultaneous on-off control of six *motion classes* was first demonstrated in the SVEN hand in the 1970's."

  The difference between the terms *motor functions* and *motion classes* is explained in Section II.A on p. 666.

Paper G Simultaneous Proportional Control of Dualfunction Upper Limb Prostheses

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