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

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Abstract—The recent introduction of novel multifunction hands as well as new control paradigms increase the demand for advanced prosthetic control systems. In this context, an unambiguous terminology and a good understanding of the nature of the control problem is important for efficient research and communication concerning the subject.

Thus, one purpose of this paper is to suggest an unambiguous taxonomy, applicable to control systems for upper limb prostheses and also to prostheses in general. A functionally partitioned model of the prosthesis control problem is also presented along with the taxonomy.

In the second half of the paper, the suggested taxonomy has been exploited in a comprehensive literature review on proportional myoelectric control of upper limb prostheses.

The review revealed that the methods for system training have not matured at the same pace as the novel multifunction prostheses and more advanced intent interpretation methods. Few publications exist regarding the choice of training method and the composition of the training data set. 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. In order to further promote the development of proportional myoelectric control, these topics need to be addressed.

Index Terms—Electromyography, estimation, proportional control, prosthetic hand, prosthetics.

I. INTRODUCTION

APPROXIMATELY 100 000 people in the USA have a major upper-limb loss, 57% of them being transradial amputees [1]–[3]. About 80% use a prosthesis [4]. Myoelectric controlled devices are used by roughly 30%–50% [3], [5].

The last decade has seen a remarkable increase in upper-limb prosthetics research; a simple online search indicates that up to 2010, roughly 4000 publications contained one of the terms *hand prosthesis* or *prosthetic hand*, 70% of which were published during the years 2001–2010 [6]. Numerous new methods and devices have been described by independent authors, causing a divergence of practice and terminology that reduces readability and complicates comparison of different contributions. This issue has previously been noted in the area of outcome measures, and efforts have been made to alleviate the problem [7], [8].

This paper presents an overview of literature related to, or relevant for, proportional myoelectric control. For the sake of readability, an unambiguous terminology is suggested for the subject based on pertinent literature. We also seek to illustrate and clarify the complexity of the subject and the relationships between different notions that are frequently confused. Several of the terms treated are equally relevant for upper-limb prosthesis control in general and proportional myoelectric systems in particular.

A. Proportional Versus On–Off Control

In order to appreciate the nature of proportional control, we first briefly consider the alternative, namely *on–off control*, also known as *bang-bang control*, *crisp control*, or *binary control*. In this control mode, 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). Even seven decades after its conception [9], the robustness and intuitiveness of this simple control mode is proven by its continuing popularity. 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. We discourage usage of the term "digital control" in a prosthetics context to avoid confusion with modern digital control systems.

We define *proportional control* as follows.

> *Definition 1: 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 negligible.

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.

A simple example of *proportional myoelectric control* is a system in which the electromyogram (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 [10]. This functionality is essentially offered by several manufacturers of commercial prostheses.

The human neuromotor system exhibits proportional control abilities according to the above definition, in the sense that we can vary joint torques, speeds, positions and contact forces continuously at will. Similar qualities can also be seen, though with inferior fidelity, in most body powered prosthesis systems, to the extent that a harness and cable physically link the movement of a body part to the movement of the prosthesis. It is therefore not surprising that the notion of proportional myoelectric control was introduced as early as in the 1950s by Battye [11], Bottomley [12]–[14] and Rothchild [15], [16].

In 1974, Roesler claimed that proportional control is required for quick grasping of objects, while at the same time having the possibility of slow and precise prehension [17]. In a renowned work shortly after, Sörbye 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 [18], [19]. Three decades later, Lovely [20] claimed that the need to control the finger speed originally arose because of the slow motors 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, 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 [21]. The controversy around the necessity and appropriateness of proportional control in upper limb prostheses thus is still very much alive.

To the author's best knowledge, the prevalence of proportional control has not been reported in the scientific or engineering literature, but it is currently (as of November 2011) available as an option from all manufacturers of commercial myoelectric prostheses; Liberating Technologies [22], Motion Control [23], Otto Bock [24], RSLSteeper [25], Shanghai Kesheng [26], and Touch Bionics [27]. In research, groups have presented several forms of proportional control, while using different names for the various concepts and methods. Thus, the purpose of this paper is to suggest a *common terminology* for all types of prosthesis control, not just proportional myoelectric control, and to utilize this terminology to summarize the methods for proportional control that have been developed and tested during the last 60 years.

Section II of this paper contains a review and recommendations regarding terminology, including a visualization of the relationship between various commonly used expressions.

Section III describes the history and the methods within proportional myoelectric control. Methods are grouped thematically but presented chronologically within each topic.

The review does not include research on *lower limb* prosthetics, which is an emerging application of myoelectric control. Also, we have reviewed only those signal processing methods that are specific to proportional control.

II. TERMINOLOGY IN PROSTHESIS CONTROL SYSTEMS

The recent introduction of novel multifunction hands as well as new control paradigms like targeted muscle reinnervation [32] and implanted electrodes [3], [33]–[35] increase the need for advanced prosthetic control systems. In this context, an unambiguous terminology and a good understanding of the nature of the control problem is important for efficient research and communication concerning the subject.

We have defined *proportional control* in the Introduction. However, sometimes relevant research is published on the topic without using that term. For example, expressions like intuitive, natural, dexterous, continuous, variable [43] or simply myoelectric control [44], [45] have been used, sometimes for proportional control and sometimes for other *control strategies* (see Table I).

TABLE I
NOMENCLATURE

| Expression	Description	Examples
Degree-of-freedom (DOF)	The number of parameters that are necessary and sufficient for a unique characterization of the kinematic configuration (geometry) of the system [36], for example independent displacements or rotations. In a prosthesis, each DOF usually corresponds to a passive, body-powered or motorized rotational joint.	Wrist pro-/supination, elbow flexion/extension, thumb flexion/extension
Motor function	In humans, this term denotes the ability to use and control muscles and movements. In a prosthesis control context we propose to use this term for any distinct type of prosthesis movement, involving a single DOF (an *elementary* motor function) or several (a *complex* motor function), without reference to speed or direction.	Elementary: Elbow flexion/extension, wrist pro-/supination. Complex: Power grip, key grip.
Multifunction device	A device that exhibits more than one motor function.	i-Limb Ultra [27], BeBionic hand [25]
User intent	A set of motor functions that the user intends to exploit, or motion classes that the user intends to activate.	Move the hand towards an object, open the hand and grasp the object.
Motion class	Similar to motor function, except that a motion class has an explicit direction or zero speed (i.e. "rest"). Thus, three motion classes crudely correspond to a single motor function. The notion of a motion class relates to a crisp classification scheme, where different classes are typically mutually exclusive.	Elementary: Elbow flexion, elbow rest, elbow extension. Complex: Power grip close, hand rest, hand open
Input signal feature	Some quantifiable property of the input signal(s), extracted in order to concentrate or isolate the essential information in the signal(s). The process of feature extraction is well known from the field of pattern recognition, with which the present definition is intended to be fully compliant. In the context of *mapping* in *proportional control*, features are often referred to as *parameter estimates*.	Zero-crossings, root-mean-square, mean absolute value, etc. (see Section III-A)
Sensor modalities	Distinct types of sensors through which a system can receive input from the environment. Different modalities imply the sensing of different quantities and/or sensing of the same quantity using different techniques.	Surface EMG electrode, contact force sensor (e.g. touch pad), accelerometer.
Multi-modal	Involving multiple sensor modalities.	Fig. 1, Example 3.
Intent interpretation	Inference about the user's intentions based on available input signals and prior knowledge. Corresponds to layers 4 & 5 in Fig. 1.	Fig. 1, Examples 1-3.
Pattern recognition	Given some examples of complex signals and the correct decisions for them, make decisions automatically for a stream of future examples [37]. This definition covers both of the terms classification and mapping, as described below.	Classification, mapping.
Classification	Assignment of each set of input feature values to one of a given set of classes. In prosthetis control, it means to assign a set of *input signal feature* values to one of a given set of *motion classes*.	Linear discriminant analysis (LDA).
Mapping	A function that maps a set of input values (signal features) to a set of output values (continuous actuator control signals which may be used for calculation of actuator setpoints). The method used for finding a suitable mapping is called *regression*, and the act of applying the mapping is called *estimation* or *prediction*.	Linear or nonlinear regressor [38].
State	The present condition of a system (e.g., a *state machine*).	"Squeeze" state in [39].
State machine	A system in which the behavior is dependent on the present state and state transitions are triggered by certain discrete events. The transition from one state to another (e.g., from *hand control* to *wrist control*) may be triggered by a timer or the input from the user (e.g., a *co-contraction*). If the states are ordered in predefined *sequences*, it can be referred to as a *sequential control system*.	Southampton Adaptive Manipulation Scheme (SAMS) [39].
Actuator control signal	The signal input to the motors in the prosthesis.	Usually a pulse-width modulated voltage signal.
Mechanical impedance control	Exhibited by a controller that attempts to implement a dynamic relation between manipulator variables such as end-point position and force rather than just control these variables alone.	[40]
Hybrid prostheses	Prostheses with some body powered components and some electric components [21].	An electric terminal device combined with a body-powered elbow
System training	Training of the prosthesis control system to recognize input signals from the prosthesis user. This is often just referred to as *training* or *supervision* in pattern recognition. Not to be confused with "User training".	Prosthesis-guided training [41], [42].
User training	Training of the user's ability to control a prosthesis. Not to be confused with "System training".	[18]
Control scheme	Equivalent to *Control strategy* (below).	
Control strategy	The terms control strategy and control scheme are applied to various pars or aspects of a prosthesis system, ranging from the input sensor configuration (e.g. single vs. two-site EMG control) and intent interpretation (e.g. proportional vs. on/off control), to the control and configuration of actuators (e.g. force vs. speed or position control). Yet other authors use these terms to denote the entire prosthesis control system, including all eight functional layers of the present model (Fig. 1). We recommend the latter interpretation, and suggest that the individual layer or layer group names in the model are used when communicating about the corresponding aspects of the control problem.	Three examples are shown in Fig. 1.
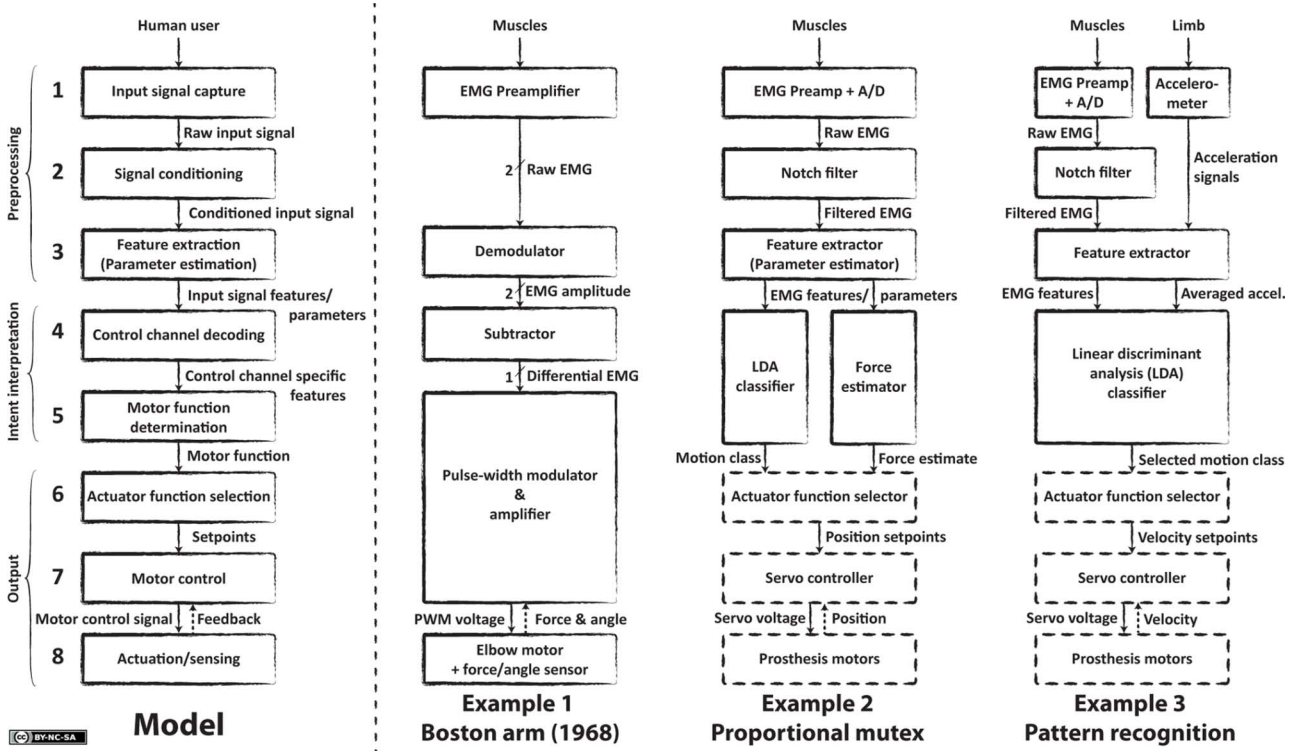
Fig. 1. A functionally partitioned model and corresponding taxonomy for the prosthesis control problem. It is an augmented version of the model proposed by Losier [28]. Three examples are given: Ex. 1 is the control system for the Boston Arm in the year 1968 [29]. Ex. 2 is a proportional mutex control system, where levels 1–5 correspond to the research by Hudgins [30] 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 Fougner [31] and the dashed lines represent a possible implementation of layers 6–8. This figure is licensed under a Creative Commons BY-NC-SA license.

TABLE II
EXAMPLES OF CLASSES, FUNCTIONS, AND DEGREES-OF-FREEDOM

Degree of freedom	Motor function	Motion class
Wrist pro-/supination	Wrist pro-/supination	Pronation
		Rest
		Supination
Thumb flexion/extension	Pinch grip/open	Pinch grip
		Rest
Index finger flexion/extension		Pinch open

A. Definitions

In Table I, we define many of the expressions commonly used and present examples of each of them. We propose to avoid the terms *control variable* and *controlled variable* (unless properly defined) since they are vague and their meaning is ambiguous across different fields.

The difference between *degrees-of-freedom*, *functions* and *classes* may not be obvious. In order to illustrate the relationship between these terms, we present Table II where typical examples are included. Some functions and classes have identical names. They may however still be different, since motor functions usually include a possibility for resting, while for classification there is an additional class called "rest."

B. Taxonomy for the Prosthesis Control Problem

In order to illustrate the relationship between some of the various terms commonly used in prosthesis control, in Fig. 1 we present a functionally partitioned model and corresponding taxonomy for the prosthesis control problem. It is an augmented version of the model proposed by Losier [28] and has eight layers.

1) **Input signal capture** is the capture of signals from the input sources, for example from EMG electrodes. In a modern prosthesis system this will normally include some preamplification and analog-to-digital (A/D) conversion.

2) **Signal conditioning** is the processing of input signals in order to get the signal ready for feature extraction. This can, for example, include notch filtering to suppress power line interference.

3) **Feature extraction** is the calculation of signal features based on the conditioned input signal. Traditionally the most common feature has been the EMG mean absolute value estimate. This layer can be called **parameter estimation** in a proportional control system.

4) **Control channel decoding** is the splitting of available input signal features into separate and preferably independently controllable signals. As an example, the implementation of the three-state UNB controller uses "level decoding" for control channel decoding of the EMG signal from a single electrode site [46].

5) **Motor function determination** is the mapping of decoded input signals to available motor functions. In the

Fig. 2. A 3-D representation of myoelectric control for upper limb prostheses. Two examples of commercial prostheses are indicated in the diagram, as well as two examples from research. The "Research state-of-the-art" example refers to a system with multi-modal pattern recognition described by Fougner [31], and the "SVEN hand" example refers to the research by Almstrom *et al.* [47], [48]. Note that the boxes on the *Motor function determination* 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"). This figure is licensed under a Creative Commons BY-NC-SA license.

implementation in contemporary systems, the functions of "Control channel decoding" and "Motor function determination" will often be implemented as a single module; for example in pattern recognition (as illustrated in Fig. 1 Ex. 2).

6) **Actuator function selection** is the translation of selected motor functions (for example "pinch grip") into setpoints (for example position setpoints) for the available actuators.

7) **Motor control** is the low-level control of the input to the motors, based on setpoints and sensed feedback from the motors. A servo controller is often responsible for this function.

8) **Actuation/sensing** is normally performed by the prosthesis motors and feedback sensors.

The layers of this model represent *principal functions* of the control system; i.e., not the physical software or hardware modules. In a physical system, the actual implementation may miss one or more of these functions, some functions may be merged in a single module, and the functions may be applied in a different order. For example, a typical commercial active EMG electrode will include both bandpass filtering, amplification and feature extraction (in the form of amplitude demodulation). Thus, the active electrode module implements all layers 1–3 in the model. 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).

Three additional terms are included in the model.

Preprocessing (Layers 1–3) is the collection of information from the human user. The physical implementation usually includes sensors and signal processing

Intent interpretation (Layers 4–5) is the interpretation of user intent from the available information (which was collected in the previous layer). It can be seen as the *essential* part of a prosthesis control system in the sense that it is this functionality that decides the high-level control properties experienced by the user.

Output (Layers 6–8) is the implementation of decisions made (in the *Intent interpretation* layer). All of these functions are often implemented near the actuators.

Our model is intended to fit all prosthesis control systems. To illustrate how they can be fit, we have presented it along with three implementation examples in Fig. 1.

C. Activation Profile

For illustration purposes we may choose to decompose this problem into three orthogonal (i.e., independent) axes as shown in Fig. 2. Any combination in the space spanned by these axes,

can be or has been used in myoelectric control of upper limb prostheses. Four existing combinations have been indicated as examples in the figure.

The axes *Preprocessing* and *Intent interpretation* have already been described as layers 1–3 and 4–5 of the model in Fig. 1. The vertical axis, *Activation profile*, is another property of the control system that is particularly relevant for this paper since it distinguishes explicitly between *proportional* control and various forms of on–off based schemes.

For *on/off control* the actuator control signal can only be on or off, i.e., a binary signal. In this way a *motor function* (e.g., hand open, grasp, or pronation) can be activated or deactivated, but the user can not control it in a proportional manner; in order to control e.g., the actual opening angle of the hand, the user must resort to switching it off at the exact right moment. This concept is widely used in today's prostheses, mainly because it has showed itself relatively robust and predictable.

Multi-level control means that the user can achieve multiple actuator control signal levels for a prosthesis function but generates a noncontinuous control signal [49], i.e., not *essentially continous* as described in Section I. It may still be continuous in *time*. An example of multi-level control is when a system can have two possible servo motor speeds, "slow" and "fast," depending on the strength of the myoelectric signal.

The *decision ramp function* was introduced by Simon [50] in order to minimize the effect of misclassifications in a pattern recognition system. Note that this method can be combined with the multi-level control.

D. Intent Interpretation

The axis called *Intent interpretation* in Fig. 2 relates to the complexity of the system's intent interpretation apparatus. Several methods can be used for this task, and a multifunction prosthesis naturally requires a more sophisticated motor function determination method than a single-function prosthesis.

1) Single Function Systems: As it may be observed in Fig. 3, most of the research on proportional control until the 1990s was for single-function systems [10], [51]–[54] (multi-function systems are marked as "proportional mutex" or "simultaneous" in the figure). For such a system, the selection of motor function is trivial. Thus, the research in this domain has usually been focused on the other parts of the system such as the feature extraction. This is presented in Section III-A.

2) State Machine: For a *state machine* (see Table I) with two or more states, the user can for example use input from other sensor modalities like force and slip sensors [39], [55]–[58] to switch between states. A special case is a *sequential control system* where the user can use *co-contractions* of antagonistic muscles [20] or a mechanical switch [59] to scroll through a sequence of available states. The transitions between states may also be triggered by a timer, as demonstrated in the ToMPAW arm [60].

3) Classification: *Pattern recognition* is a popular term in prosthetics research. The expression has so far mainly been used for *mutex* (mutually exclusive) classification, i.e., selection of a single intended motor function without having the possibility of *simultaneous* control of multiple functions. The first well-

known example among many publications is the method developed by Hudgins [30] and Englehart [61] which was implemented and used by Lock and Scheme [62]. Since this paper is focusing on proportional control we will not go into details on classification algorithms.

A few of the publications on classification have included proportional control [30], [49], [63]–[66] (see Section III-B).

4) Simultaneous Control: Simultaneous *on–off control* of six motor functions was first demonstrated in the *SVEN hand* in the 1970s [47]. The results of the clinical trials were promising although the hand was not reliable (nor portable) enough for testing outside of the laboratories [48]. This was one of the first examples of pattern recognition in myoelectric control for prostheses.

For a simultaneous *proportional* myoelectric control system, which has been approached a few times the last five years with different mapping functions [67]–[75], some or all of the available functions will be controlled simultaneously. Many of the researchers have been inspired by other fields of study like kinesiology [76]–[78] and tool ergonomics [79]–[81] where similar methods are needed for force estimation or motion prediction. While the results have been promising, the methods have not yet arrived in commercial prostheses.

E. Body-Powered Prostheses and Extended Physiological Proprioception

Most, if not all, body-powered/cable-driven prostheses inherently have proportional control, since there is a direct mechanical coupling between the user's body and the actuated joint, unless the cable is just used for controlling a switch. This mechanical coupling offers feedback to the user—a concept introduced as *extended physiological proprioception (E.P.P.)* by Simpson in the 1970s [82]–[85]. It works basically the same way as a blind person's cane which makes one able to sense the surroundings.

E.P.P. may also be used in *powered prostheses*, as described by Weir [86]. *Muscle tunnel cineplasty* or *tendon exteriorization cineplasty* can be used for interface with the muscle and will offer a one-to-one relationship between position, speed and force of the controlling muscle and that of the prosthetic component. In other words, it will offer proportional *mechanical* control and E.P.P.

Other examples of proportional mechanical control are those described by Salisbury and Mortimer [87], [88]. We will not go into details on these since the focus of this part of the paper is on proportional *myoelectric control*.

III. REVIEW OF PROPORTIONAL MYOELECTRIC CONTROL OF UPPER LIMB PROSTHESES

Several authors have performed reviews of powered upper limb prosthesis research, often with a smaller portion of proportional myoelectric control [30], [89]–[95]. The history of research on proportional myoelectric control is presented in Fig. 3 in order to illustrate the relationship between the publications and their inspiration. Publications on lower limb prostheses are omitted. Some related research commonly referred to is also included, such as research from *ergonomic tool design* and proportional control with other inputs than EMG.
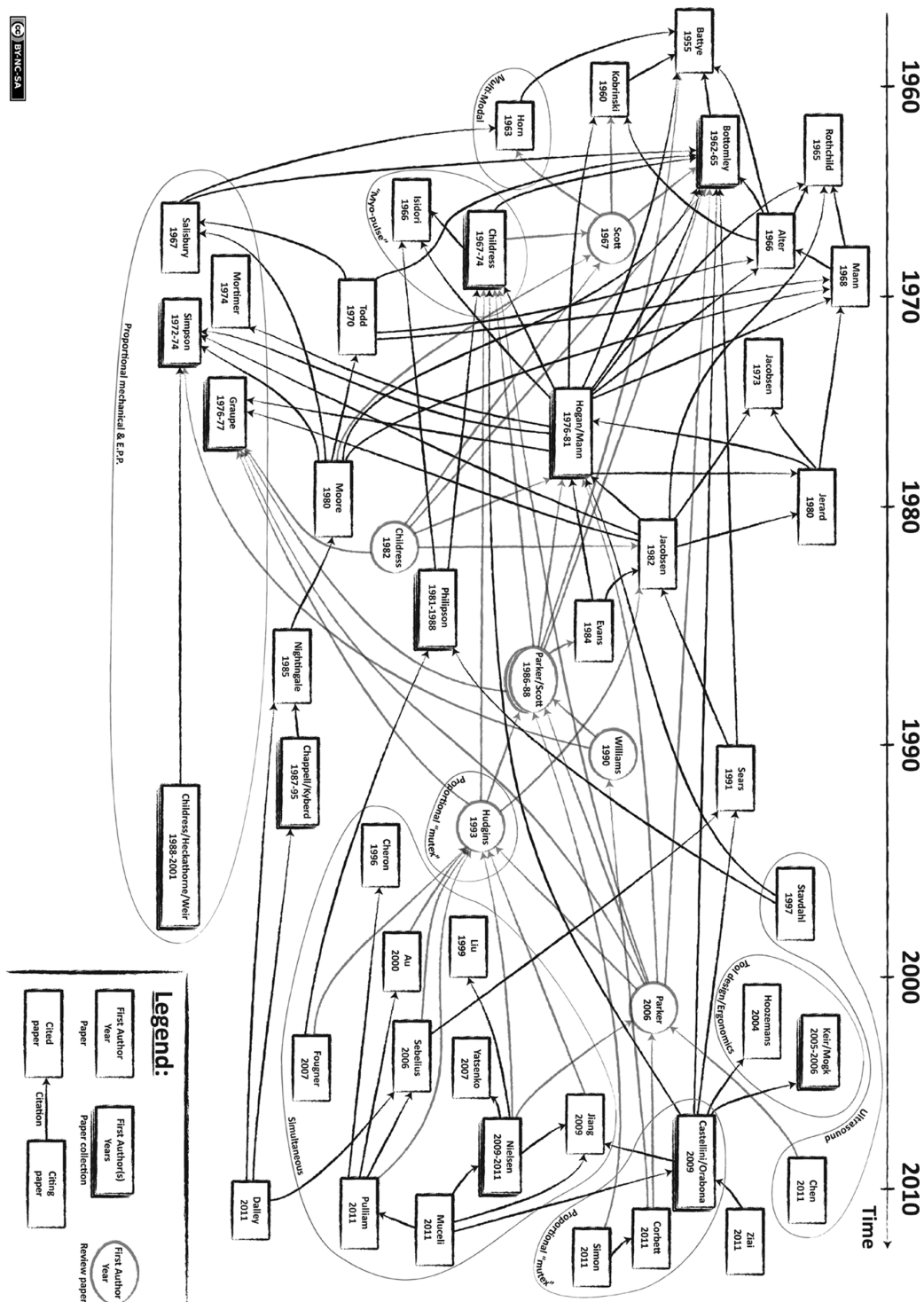
Fig. 3. A chronological representation of papers on myoelectric proportional control for upper limb prostheses. Note that usually only the first author is mentioned in the figure—while all other authors are included in the References section. Papers that only refer to other papers on proportional control, usually in a review, without actually using proportional control, are shown as grey "nodes." The arrows represent citations; pointing from the citing paper towards the cited paper. This figure is licensed under a Creative Commons BY-NC-SA license.

A. Parameter Estimation

This section of the paper will describe the various signal features (sometimes called *parameter estimates*) that have been used for proportional control. Most of them have an abbreviation, as presented in Table III.

1) EMG Parameter Estimates: Battye originally proposed to use EMG for proportional control [11]. **Mean absolute value (MAV)** was the feature used by Bottomley [12]–[14] when he

suggested a solution in 1963. He described proportional control of a force- and velocity-feedback powered hook, allowing for speed control when moving freely and force control when gripping. The feature used was an amplified, rectified and smoothed electrode signal. The MAV from two electrode pairs were compared and the differential signal was used to open and close the prosthesis hand. To attenuate output noise, a backlash generator was introduced. This solution was adapted by Horn [96]

TABLE III
ABBREVIATIONS FOR EMG PARAMETERS/FEATURES

Abbreviation	Name
MAV	Mean absolute value
MSV	Mean square value
MYOP	Myo-pulse
NT	Number of turns
RMS	Root-mean square
SSC	Slope sign changes
WAMP	Willison amplitude
WF	Windowed Fourier transform
WL	Waveform length
WPT	Wavelet packet transform
WT	Wavelet transform
ZC	Zero-crossings

in a *multi-modal* version (see Section III-A2). Independently of Battye and Bottomley, Rothchild in 1965 developed a proportional control system for the Boston Arm [15], [16], a solution further developed by Alter and Mann [29], [97], [98]. Alter did not find Bottomley's backlash element to be useful, and Hogan [99] later proposed that the main problem of the backlash element was the time delay that it introduced to the system. Bottomley's solution was used for a hand and the Boston arm control system (illustrated in Fig. 1 Ex. 1) was used for an elbow, but in most other respects the systems were almost identical.

In 1966, Isidori and Nicolò [45], [100] created an instrument to calculate a feature of the EMG that was later used and referred to as **myo-pulse (MYOP)** by Childress [101]–[103] and Philipson [104], [105]. Isidori emphasized that the myopulse feature was almost linearly related to muscle force. This feature is a weigthed mean of the time during which the EMG signal is above a specified threshold, and it can be used directly as input signal to a motor drive in a prosthesis, in a way similar to "pulse-width modulated" (PWM) control.

Parker [46] and Hogan [106], [107] developed two very similar *model-based* formulas for an "optimal myoprocessor" (using EMG to estimate muscle force). Hogan and Mann later used it for proportional *mechanical impedance control* [40], [108] (see Section III-E). The optimal myoprocessor happens to be the feature commonly called **root mean square (RMS)**.

Another model-based approach was demonstrated by Evans [109] when he used a **logarithmic nonlinearity** in an EMG force estimate. This allowed him to map the EMG into an additive "control signal-plus-noise domain" so that he could use a Kalman filter as a muscle force estimator.

In the 1980s, several EMG features were compared by Philipson for isometric force estimation [110]. He tested **zero-crossings (ZC)**, **Number of turns (NT)**, and **Willison amplitude (WAMP)** along with three of the features previously mentioned (MAV, RMS, MYOP). *Linearity* was one of the characterizations he tested. An important result was that MAV performed almost as well as RMS.

Hudgins [30] proposed a feature set consisting of MAV, ZC, **slope sign changes (SSC)**, and **waveform length (WL)**. It was used for classification and has become known as the "TD (time-domain) feature set," but it was also suggested to use it for proportional mutex control (see Section III-B). In addition to the TD set, Englehart [61] tested a few time-frequency-domain sets (WF, WT, and WPT). It was shown that the TD feature set

outperformed the time-frequency-domain sets [111]. However, they were only tested in classification.

Jiang used **mean square value (MSV)**, a parameter related to MAV and RMS by the square-root, for force estimation [70].

Combinations of up to four out of 15 different EMG features (including most of those previously mentioned) were assessed by Fougner [73] for joint angle estimation. It appeared that the best feature set was specific to each user.

2) Other Sensor Modalities and Multi-Modal Approaches: Several other modalities than EMG can be exploited for prosthesis control. Among the relevant sensor modalities, some have been used in combination with EMG (in a multi-modal approach) and some are used alone. One may also combine *body-powered* solutions with myoelectric control to be able to control more than one function [112]. However, these prostheses are usually referred to as *hybrid*, not multi-modal, as long as the two input sources are used for separate functions [21].

The following list includes modalities that have been used in relation to prosthesis control, i.e., in products or in research projects. We have also included some modalities which have only been used for classification.

Potentiometers and joysticks have been popular. Actually, a potentiometer was used in a *multi-modal approach* by Horn already in 1963, when he measured rotation of the stump with a potentiometer and used this signal along with EMG to control a prosthesis proportionally [96]. Graupe presented a system with one electrode to select motor function and measurements of toe movement to control the strength of prehension, speed or torque [113]. Swain used a joystick along with EMG for his version of the SAMS system [114]. More recently, a joystick was used to measure shoulder position and combined with EMG by Losier [115]. Another example is the recent Luke Arm patent [116], where a foot joystick is used for endpoint control.

Touchpads and force sensors have been popular among manufacturers and users, as an alternative to myoelectric control for patients with lack of good EMG sites. They can for example be installed in a harness at the shoulder and offer proportional control.

Force measurements on the EMG electrodes were used to measure external forces in a multi-modal approach by Fougner and Stavdahl [117]–[119].

Mechanomyogram was used by Silva [120] for classification. This modality has also been referred to as phonomyogram, vibromyogram, soundmyogram, or acoustomyogram.

Myokinemetric signal, also called muscle bulge, was used by Kenney [121] for classification.

Accelerometers have been used to measure limb position in multi-modal classification by Fougner [31].

Ultrasound was used by Stavdahl [122] for force estimation and Chen [123] for angle estimation.

Force- and slip-sensors were used by Todd [124], Moore [125], Nightingale [55], Chappell [56] and Kyberd [39], [57], [58] in the Southampton hand. This system was a *state machine* (see Section II-D2) in which two of the available states had *proportional* control and the mechanical inputs triggered state changes. In this example, the sensors were used in Layer 8 (Actuation/sensing) of the model in Fig. 1, as opposed to the

previous examples where the sensors are used in Layer 1 (Input signal capture).

B. Intent Interpretation in Multifunction Systems

Most of the recent research on proportional myoelectric control has been on *multifunction* systems. A natural consequence is that the research has focused on the *intent interpretation* part of the system more than previously. The methods used for intent interpretation in these systems can be divided into two types.

Proportional mutex is when the system consists of both a *mutex classifier* (such as an LDA classifier) for motor function determination, and some sort of *mapping function* (also called *regression*, *estimation*, or *prediction*; see Table I) to control the selected motor function in a proportional way. In that way, the system is able to exhibit proportional control of multiple functions, but only for one motor function at the time [30], [49], [63]–[66], [126]. One example of such a control system is illustrated in Fig. 1 Ex. 2.

Simultaneous proportional control is when several motor functions can be controlled simultaneously and proportionally. All but two of the publications so far have used some type of *artificial neural network* (ANN) for the mapping from input signal features to motor function. These have been multi-layer perceptron (MLP) networks [38], [67], [68], [71]–[75], [77], [78], [127], or a recurrent neural network [76]. Most of these examples were force/torque estimation; the others were position/angle estimation [68], [73].

The other two solutions presented are the EMG energy orthonormalization along principle movement vectors, by Yatsenko [69], and the DOF-wise nonnegative matrix factorization, by Jiang [70]. The strength of Jiang's method is the semi-unsupervised nature, i.e., that it does not require force measurements in the training.

In addition, a linear mapping function was presented by Fougner [68], [73]. It was tested for angle estimation and compared with an MLP network. The much simpler linear mapping function was almost as good as the MLP network. A similar solution (referred to as ordinary least squares linear regression) has also been tested for torque estimation by Ziai [38] along with ANN, support vector machines (SVM), locally weighted projection regression (LWPR), and a physiological based model (PBM). The linear mapping was shown to have short training time and good results compared to the other more complex estimators.

The process of finding a mapping is sometimes referred to as *regression* [38], [64], [65], [79], [80], [123], and the mapping itself has been given various names like force estimation [38], [64], [79]–[81], trajectory estimation [76], force prediction [64], or motion prediction [78]. However, the purpose is the same: mapping of input signal features to motor function.

C. Training

All prosthesis control systems need to be adapted to the human user. We choose to name this adaptation *system training* and it must not be confused with *user training* (see Table I).

System training methods are not specific to proportional control, but we will give examples of how these methods have been

used in proportional control. The choice of *training data* will be discussed Section III-D.

1) Gain and Treshold Adjustments: For systems with few EMG electrodes and few motor functions, the training has traditionally been manual adjustments of *gain* in the preamplifier and *thresholds* for activation of motor functions [11]–[14], [44], [45], [96], [97]. This process is still commonly used in many commercial upper limb prostheses and may often be referred to as "tuning of parameters," "system adjustments," or "adaptation to the user" in the research literature. Such manual methods become increasingly impractical as the number of parameters increases.

2) Tracking and Computer-Guided Training: Corbett [66] has presented a training method based on *tracking*: Users where instructed to trace a cursor along a target waveform on a computer screen. This method has been used by Simon [49] for multiple DOFs, training each DOF separately.

An alternative method is to track the motions of an animated or video-recorded hand on the computer screen [63], [70], [126].

3) Bilateral Training (Mirroring): Asking a unilateral prosthesis user to "mirror" the contralateral hand's motions with the phantom limb is an efficient way to get a measurable reference for the training, because it makes it possible to record the intended movements. Nielsen measured the muscle force on the contralateral hand and used it for proportional force estimation [72]. Several research groups have measured the movements of the contralateral hand with a *data glove* and used it for proportional control [67] or on–off control [128]. Another group recorded angles of the controlateral hand with a camera system and used it for angle estimation [127]. Mirroring has also been used for both proportional and on–off control of an orthosis [43].

4) Prosthesis-Guided Training: This revolutionary training method was introduced by Lock [41] and Simon [42]. It was presented for crisp control with pattern recognition, but it may also be used in the case of proportional control. The procedure is simple: 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 *retrain* whenever needed, just by pushing a button and thereby starting a training procedure.

A drawback is that this method can not be used to train a grip force estimator for proportional control, unless a special-made solution is created for it.

D. Composition of the Training Data Set

An important part of training a prosthesis control system is to compose the training data set in an appropriate way. We have seen that the composition of training data is very important for robustness in the crisp classification version of pattern recognition [31], [129]. There is no reason to believe that training data are less important in the case of proportional control.

Naturally, a training set for proportional control should contain *continuous movements*, i.e., not only resting and maximum contractions. If trained only with on/off-movements, the system might end up implementing on–off control.

For a system with simultaneous proportional control, the training data needs to contain *s*imultaneous movements, unless some kind of *interpolation method* is being used [130].
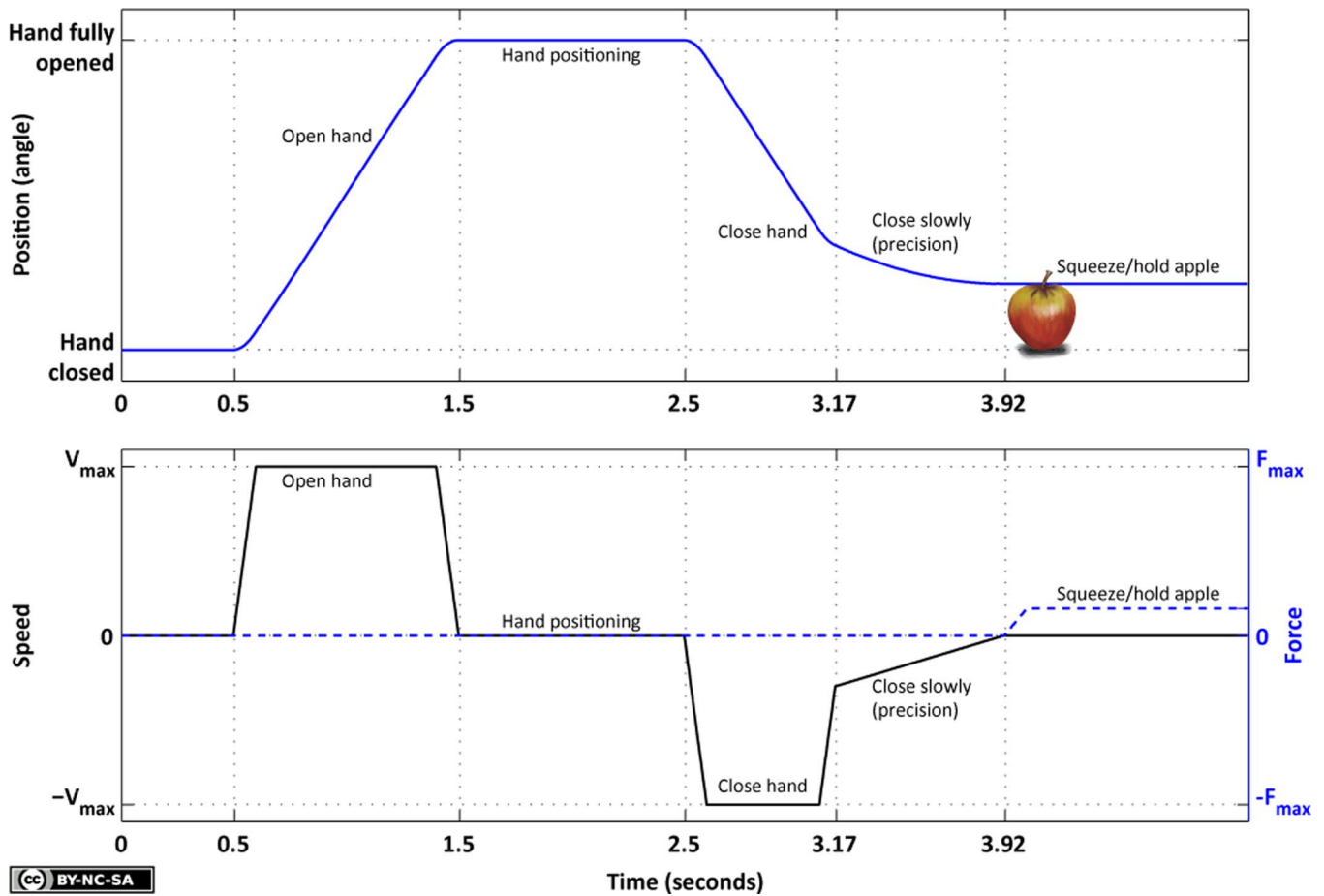
Fig. 4. Position (top) and speed/force (bottom) for a prosthesis hand grasping an apple. The simulated data are an example but illustrate a realistic scenario. This figure is licensed under a Creative Commons BY-NC-SA license.

Both of these statements are related to the fact that *the training set needs to be as realistic as possible*.

E. Choice of Controlled Actuator States

The choice of *actuator control signal* and *controlled actuator state* does not need to be identical. If we estimate forces, we may still use those estimates to control velocities of a servo motor, or vice versa (as illustrated in Ex. 1 of Fig. 1). Velocity, force, position or any *combination* of these variables can be controlled in a proportional manner. Since they are all interconnected, you may estimate one of them and calculate the other. Actually, you may achieve the exact same motion by controlling the position in a smooth way as by controlling the speed in an on–off manner. As an example, Fig. 4 illustrates a scenario of opening the hand and grasping an apple, when using a hand prosthesis. It may be important to have slow or precise control just prior to grasping the apple. In this example it was achieved by using two available accelerations (fast or slow), but proportional control of position or speed could give the same effect.

Hogan proposed to control the *dynamic relation* between manipulator variables, such as end-point position and force, rather than just control those variables alone [40], [108]. This is relevant for prostheses just as much as for robot manipulators. The relation is called *mechanical impedance* and illustrates that we do not need to choose between position control or force control.

Thus, the most important part of this problem is to find a reference parameter that can be measured during the training (tuning) process. Estimated parameter values can then be used to control any output variables of the prosthesis: Choose the ones that *works better*. Note however that *work better* implies choice of an outcome measure, which is a challenging topic by itself [7], [8], as seen in (Section III-F1).

F. Outcome Measures

1) Performance Evaluation During Training: Performance evaluation is an important part of the training. Very often, the optimization criterion in the training method is to reduce some kind of error measure. However, the RMS error which is most frequently used, is not necessarily a good measure of performance for a prosthesis control system. This has been shown by Hargrove [131] and Lock [63] for crisp classification, and by Fougner [73] for simultaneous proportional control.

This problem is illustrated in Fig. 5, where RMS error does not work well as a performance measure. We have compared two possible joint angle estimates with an intended joint angle. Estimate 1 is just the mean joint angle (dashed–dotted), i.e., at a fixed joint angle, while estimate 2 (dashed) has approximately the correct shape but contains an offset.

Thus, when the training method is based on minimization of the RMS error, optimal results will not necessarily be achieved.
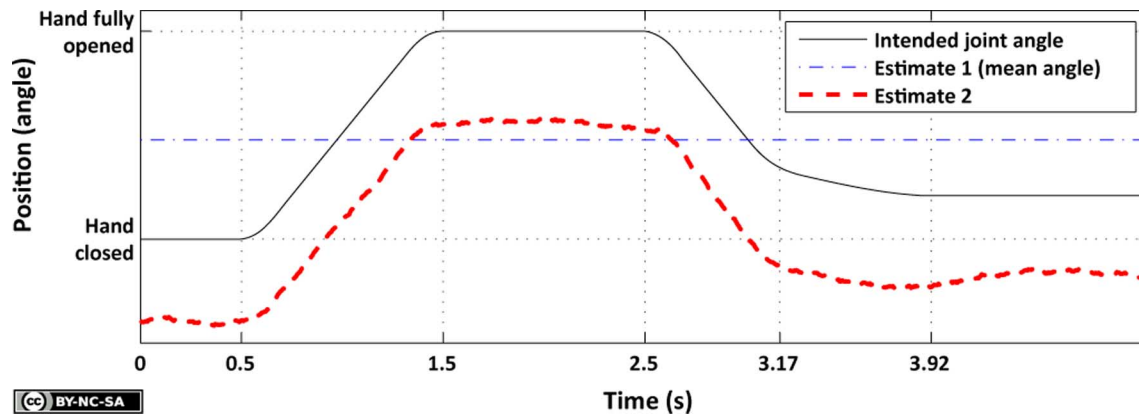
Fig. 5. Illustration of estimation problems when RMS error is used as performance measure. The data are generated as an example. This figure is licensed under a Creative Commons BY-NC-SA license.

2) Outcomes After Training: In order to find the best control method, in general or for a specific prosthesis user, the outcome needs to be evaluated in the *Function*, *Activity*, and *Participation* domains [7], [8]. A method which works well in the *Function* domain may perform worse in the *Activity* domain [63], [131] or in the *Participation* domain—and vice versa. Thus, in principle, a control strategy should not unconditionally be rejected just because it performs poorly in a functional test.

G. Sensory Feedback

Since the use of feedback in powered upper limb prostheses is not specifically related to proportional control, we will not go into details. The review papers by Childress [132] and Scott [133], [134] cover this topic up to the 1990s. Sensory feedback was tried already by Kobrinski [44] in 1960 by using vibrations and sounds proportional to the squeezing force of the prosthetic hand. Although not emphasized in the description of the system, they also used proportional velocity/force control. A similar feedback solution was described by Mann [29] a few years later.

The recent introduction of targeted muscle reinnervation by Kuiken *et al.* [32] has shown promising results for sensation at the reinnervated skin [135], [136]. This can be exploited for sensory feedback from a prosthesis.

IV. DISCUSSION/PROSPECTIVES

The present paper started out as a review of proportional myoelectric control. However, during the process of collecting and interpreting information, it appeared to be impossible to give a full overview of the literature without first clarifying the terminology.

Thus, one purpose of this paper was to suggest an unambiguous taxonomy for the upper limb prosthesis control problem, which we believe is applicable also to prosthesis control in general. We emphasize that the suggested terminology is not the only possible choice, and it is not necessarily complete. Difficult choices have been made in the struggle to avoid confusing terms. We attempted to include all of the existing terms in the suggested terminology, but in cases where expressions have been used in confusing ways or in ways conflicting with other professional fields, we have introduced new terms or redefined existing terms for clarification.

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.

The literature on proportional myoelectric control has been reviewed, from the first publications in the 1960s until today.

With the recent introduction of multifunction devices and more advanced intent interpretation, the methods for *system training* are relatively immature. We have discovered that few publications exist regarding the choice of training method and the composition of the training data set. These holes will need to be filled, in order to achieve useful results with proportional myoelectric control.

During system training, outcome measures have been identified as a challenging topic. All system training methods involve some kind of optimization, and it is therefore important to find a suitable optimization criterion in order to achieve good results. Future research will need to address this topic.

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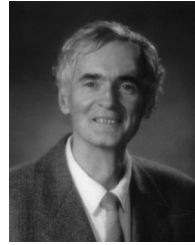
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