Study of Adaptation in Amputees

by

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Abstract

A large fraction of above-elbow amputees choose not to use available "high-tech" motorized prostheses in favor of simpler body-powered prostheses. The goal of this thesis is to begin the process of finding which aspects of body-powered prostheses make them more popular than motorized prostheses. It is desirable to objectively compare different prostheses in order to test hypotheses about which aspects of a prosthesis most affect its performance. A comparison method is proposed based on an amputee's ability to adapt to changes in a familiar task while wearing different prostheses. Crank turning is examined as a task in which adaptation can occur.

One non-amputee and one amputee subject were tested in a pilot experiment of a crank turning task. Each was allowed to become familiar with a basic crank turning task. A perturbation, in the form of an external, position-dependent torque about the crank axis, was then added. An exponential curve, asymptotically approaching a steady-state level of performance, is fit to the data that was taken during the perturbation. The amputee subject showed significant adaptation while the non-amputee subject showed no clear adapdation pattern.

Thesis Supervisor: Neville Hogan
Title: Professor

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Chapter 1

Introduction

1.1 Motivation

There is evidence that existing prosthesis technology does not meet the needs of many above-elbow amputees. Of the roughly 100,000 upper-extremity amputees that lived in the United States in 1977, it is estimated that only half of them used any prosthesis at all[10]. For one reason or another, about 50,000 upper-extremity amputees chose not to wear a prosthesis. Prosthesis technology has not changed significantly since then. Evidently, currently available prostheses are not acceptable to a significant fraction of the amputee population.

In addition, 1991 statistics show that of those that do wear a prosthesis, 90% wear a body-powered prosthesis (a simple design developed shortly after World War II) and only 10% wear a motorized prosthesis (a more sophisticated design developed in the 1960's)[8]. The fact that the more recently developed and higher technology prostheses are so much less popular may indicate that prosthesis designers do not really know how to make a "better" prosthesis. It is highly unlikely that the current state-of-the-art in body-powered prostheses represents the ideal and cannot be improved upon. It is more likely that despite their best efforts, prosthesis designers simply did not know which aspects of a prosthesis make it more popular with amputees.

The work described in this thesis is directed only at amputation prostheses for above-elbow amputees. Unless otherwise stated the term "prosthesis" should be interpreted as an above-elbow arm prosthesis. Various prosthetic wrists and terminal-devices are not discussed here. Likewise, the term "amputee" should be taken to mean "above-elbow amputee".

The goal of this thesis is to take the first steps toward answering the question, "Which aspects of a body-powered prosthesis make it more popular than a motorized prosthesis?" Investigation of this question should bring to light some of the key issues in above-elbow prosthesis design.

1.2 Background

1.2.1 Development of Above-Elbow Prostheses

A detailed history of the development of above-elbow prostheses will not be presented here; however, if the reader finds a historical perspective helpful, such a history can be found in section 1.2 of Joseph Doeringer's Masters Thesis[3]. It describes the development of the two general classes of above-elbow prosthesis available today: the body-powered prosthesis and the motorized prosthesis. These two types of prosthesis will be described in more detail below.

1.2.2 Functional Description of Above-Elbow Prostheses

Body-Powered Prostheses

Body-powered prostheses are so named because the power to operate them is supplied by the amputee. A body-powered prosthesis consists of four main components: a socket, a forearm, a harness, and a control cable.

The harness fits over the amputee's shoulders and supports the weight of the prosthesis, and also provides a fixed reference point to which the control cable can be anchored. The socket fits snugly over the amputated limb and is held in place by the harness. The forearm is attached to the end of the socket by a hinge, which acts as an elbow.

The control cable ties the harness, socket, and forearm together and provides the kinematic constraint that allows the amputee to flex the prosthesis' elbow. The cable is attached to the harness near the non-amputated shoulder. It runs across the amputee's back and through a housing that is attached to the socket, and is then attached to the forearm.

The amputee flexes the prosthesis' elbow by moving the amputated limb and socket relative to the non-amputated shoulder. This is done by a combination of humeral flexion (the motion of raising the arm from the downward position until it is pointing forward) and biscapular abduction (the motion of rounding the shoulders).

The prosthesis also has an elbow lock that the amputee can activate (and deactivate) by quickly shrugging the amputated shoulder. This allows the amputee to lock the elbow at one of a range of angles–a feature that proves useful in tasks that require fine position control or the production of high contact forces. The body-powered prosthesis' ability to modulate its output impedance in this way is one of the most significant differences between it and a motorized prosthesis.

Motorized Prostheses

Motorized prostheses earn their name by being powered by an electric motor rather than by the amputee. Like body-powered prostheses, motorized prostheses also make use of a shoulder harness that bears the weight of the prosthesis, but motorized prostheses differ in that the power to produce elbow flexion is supplied by a battery, rather than by the amputee pulling on a control cable.

An amputee controls a motorized prosthesis by one of two different methods: modulating the myoelectric activity[1] in a pair of antagonist muscles (e.g.: remnants of the biceps and triceps on the amputated limb–the most commonly used antagonist pair) or activating mechanical switches using body motion. Generally, if the amputee is able to produce a sufficient control signal at the biceps and triceps, a myoelectrically controlled prosthesis is used.[11]

Two popular motorized prostheses currently on the market are the Boston Elbow and the NYU/Hosmer Elbow. Each interprets myoelectric activity differently. The NYU/Hosmer elbow monitors the difference between myoelectric activity levels in the biceps and triceps. If the difference becomes greater than a predetermined threshold, the prosthesis either flexes or extends, depending on whether the biceps or triceps are more highly activated. (Biceps activation corresponds to flexion and triceps activation to extension.) In the NYU/Hosmer elbow, all elbow motion occurs at a fixed speed.

The Boston Elbow operates similarly to the NYU/Hosmer elbow, except that it flexes or extends the elbow at a speed proportional to the difference in biceps and triceps myoelectric activity. This allows the amputee to control not only the direction of elbow motion, but the magnitude of its velocity as well.

Both of these prostheses have a very high output impedance. A motorized prosthesis' drivetrain consists of an electric motor and a series of speed-reduction gears. This causes the drivetrain to be non-backdrivable, and as a result, the elbow joint is insensitive to external torques. In other words, the prosthesis has a very high mechanical output impedance.

[1]Myoelectric activity is the electrical activity in a muscle that is present when the muscle is active. The level of myoelectric activity in a muscle roughly corresponds to the level of muscle activation.

1.3 Summary of Remaining Chapters

Despite the fact that motorized prostheses offer several benefits that body-powered prostheses do not, the latter remains far more commonly used. The remaining chapters in this thesis will present several reasons why that may be, and begin an investigation that is intended to pin down the factor or factors that make it so.

1.3.1 Chapter 2: Hypothesis Development

Chapter 2 is a summary of the background thinking and planning that have shaped the work described in this thesis. In it are proposed possible answers to the question, "Why are body-powered prostheses more popular than motorized prostheses?" One of these possible answers, the transmission of force information through the prosthesis control cable, is accepted as a working hypothesis, and a method for testing the hypothesis is discussed.

The proposed test is to have an amputee evaluate the performance of two prostheses which are identical in every respect except for the parameter of interest. This would be realized by using a prosthesis emulator as a test platform and simply modifying the level of power-assist (and thus the level of force transmitted through the cable).

The chapter also contains a discussion of the significance of force transmission in powered tools.

1.3.2 Chapter 3: Evaluating Prosthesis Performance

Chapter 3 develops in more detail a method for conducting the test proposed in the previous chapter. A task (crank turning) and perturbation (a position-dependent torque) combination are chosen as candidate for inducing adaptation in amputees.

1.3.3 Chapter 4: Hardware

Chapter 4 describes the hardware used to perform the experiment. The emulator is discussed, with emphasis on its power-assist capability. The crank and the lobe used to produce the perturbation are also described. The chapter discusses the role of the computer in the experiment. The apparatus' safety features are also described.

1.3.4 Chapter 5: Experimental Procedure

Chapter 5 contains the protocol for how the experiments were performed. It describes what happened on a given trial, how many trials the subject had under which conditions, and lists the variables that were recorded in the data files.

1.3.5 Chapter 6: Results and Analysis

Chapter 6 presents and analyzes the results of the experiments described in the previous chapter. It shows examples of typical trials, describes all the data processing that took place, presents subjects' performance levels over time, and fits an adaptation model to the data.

1.3.6 Chapter 7: Conclusions

Chapter 7 contains a summary of the conclusions drawn from the experiment and outlines plans for future work.

Chapter 2

Hypothesis Development

The research question motivating this work is somewhat general. This chapter narrows down the broad scope of the question to a hypothesis that can be feasibly tested.

The "big picture" question that this work is trying to answer is, "Why are body-powered prostheses more popular than motorized prostheses?" However, it is likely that investigation of this question will give insights into the fundamental issues of prosthesis design, regardless of whether a satisfactory answer to the question is found. The understanding provided by these insights is the ultimate goal of this work.

2.1 Strengths of Motorized Prostheses

As discussed in the previous chapter, there is no immediately obvious answer to this question. In fact there are factors that that would appear to make motorized prostheses more popular. Two of these are listed below.

2.1.1 Comfort

A limitation of body-powered prostheses is that, when the elbow is unlocked, they require large actuation forces on the cable to lift any significant load. There is relatively little travel in the physiological motions used by amputees to control their body-powered prostheses. Because of this, amputees must work at a large mechanical disadvantage in order flex the elbow through its full range of motion. The prosthesis acts as a lever that scales down the force transmitted through it by the same factor that it scales up the displacement. Inevitable friction in the cable actuation system exacerbates this problem.

Motorized prostheses are not subject to this limitation. Electric motors driving the elbow through speed-reduction gears allow the user to achieve high elbow torques much more comfortably than would be possible with a body-powered prosthesis. For example, the Boston Arm is advertised as generating 10 ft-lbs of torque at the elbow and being able to lift 9 lbs with a terminal device 14 inches from the elbow.[13] Although forces this high are sometimes achievable with body-powered prostheses, it is only at the cost of severe discomfort. Because motorized prostheses do not require the large actuation forces to lift loads, there is much less pressure on the amputee from the harness straps. In addition, adverse affects of repeated asymmetric use of the muscles in the shoulder such as scoliosis may be avoided in motorized prostheses.

It may be for these reasons that amputees using body-powered prostheses appear to use them almost exclusively in the locked position when performing any task that requires significant elbow torque production.[3]

2.1.2 Appearance

The appearance (also referred to as cosmesis) of a prosthesis may be an important factor in some amputees' decisions. Myoelectrically controlled motorized prostheses offer the cosmetic advantage of not requiring unsightly shoulder contortions to operate. The shoulder rounding required to operate a body-powered prosthesis, when combined with the quick jerks of the shoulder required to lock the elbow, can be very uncosmetic.

In addition to this, motorized prostheses do not require a cable to run across the amputee's back to the prosthetic elbow. Whether this cable is routed over or under clothing, it can be very noticeable. Both of these factors would seem to make motorized prostheses preferable to body-powered prostheses.

2.1.3 Function

Motorized prostheses allow individual degrees of freedom (e.g. elbow flexion and terminal device prehension) to be operated independently, and, in theory, simultaneously. Body-powered prostheses, however, do not.

Pulling on the control cable of a body-powered prostheses controls both the elbow flexion and the motion of the terminal device. When there is no load, the force of the spring or elastic band on the terminal device prevents it from opening[1]. Then, when use of the terminal device is desired, locking the elbow allows the cable motion to operate the terminal device. However, when lifting a load with the arm, the terminal device will open. The additional tension in the cable required to produce the elbow torque becomes enough to overcome the tension in the terminal device's elastic band or spring. In this case the two degrees of freedom cannot be operated independently. This feature of motorized prostheses would appear to give it a functional advantage over body-powered prostheses.

[1] There are both voluntary opening and voluntary closing terminal devices. In a voluntary opening terminal device, the device remains closed until the cable is actuated. This is the type of device described in the text. In a voluntary closing terminal device, the device remains open until the cable is actuated. In either case, elbow torque cannot be generated by the cable without either increasing or decreasing grip strength

2.2 Possible Factors In the Popularity of Body-Powered Prostheses

Despite the advantages offered by motorized prostheses, there are many possible reasons for body-powered prostheses' relative popularity. A few of these are listed below with brief explanations.

2.2.1 Weight

A body-powered prosthesis' simple design allows it to be much lighter than a motorized prosthesis. For example one model of body-powered prosthesis used has a mass of 912 g while a common type of motorized prosthesis has a mass of 1639 g (harness not included in both cases). When one considers the discomfort of bearing the additional weight on a consistent basis, it becomes clear that weight may be a significant factor in body-powered prostheses' popularity.

2.2.2 Market Inertia

Due to the fact that body-powered prostheses were the first to be developed and refined, they may have become a barrier to motorized prostheses entering the market. Prosthetists' and amputees' preferences may have been shaped by their initial exposure to body-powered prostheses.

However, it is important to note that motorized prostheses have been commercially available for over thirty years. It seems unlikely that they would still be so uncommon, were there not other reasons for them to be so poorly accepted.

2.2.3 Cost

Due to its simpler design, a body-powered prosthesis is far less expensive than a motorized prosthesis. For the amputee who took part in this experiment, it cost $7,000 to be fitted with a body-powered prosthesis, where it would have cost more than $30,000 to be fitted with a motorized prosthesis. This person does not have any insurance to help cover costs, so price was a very significant factor. Even when insurance companies are involved, the co-payments for a motorized prosthesis can be much larger than those for a body-powered prosthesis[9].

2.2.4 Output Impedance Toggling

Activating and deactivating the elbow lock on a body-powered prosthesis allows the user to modulate the prosthesis' output impedance. The elbow can be made to appear alternately very stiff or very compliant. This is in contrast to motorized prostheses, which are very stiff when in operation (although several types have a manually operated free-swing mode available when the arm is not being flexed or extended).

To toggle between different levels of output impedance can be helpful, depending upon the task at hand. Shaking hands with another person requires relatively low output impedance, but holding a vibrating power tool in place requires a very high output impedance. Constrained motion tasks, such as opening a door or turning a crank, generally require a low output impedance if contact forces are to be kept low. A body-powered prosthesis' ability to toggle between levels of output impedance may make it a more widely useful tool.

2.2.5 Elbow Angle Perception

In the absence of visual or other peripheral feedback, a motorized prosthesis does not provide any direct means for its wearer to sense the elbow angle. In a body-powered prosthesis, however, the position of the prosthesis socket relative to the amputee's shoulders corresponds to the prosthesis' elbow angle. Based on the position of the amputated limb (and thus of the socket) the amputee should theoretically be able to perceive the position of the prosthesis, even in the absence of visual feedback. This ability to sense prosthesis position could be a factor that makes body-powered prostheses preferable.

2.2.6 Elbow Torque Perception

A motorized prosthesis also does not provide any means for its wearer to directly sense the elbow torque. In contrast, the wearer of a body-powered prosthesis can, by feeling how hard they are pulling on the cable, sense the torque in the prosthesis' elbow. This ability to feel how hard the elbow is trying to flex may be useful in any task where force levels need to be regulated. It may also allow one to feel the environment better, and gain a better mental model of the mass, compliance, and friction that exists in whatever system one may be interacting with.

Of all the possible factors listed, it is not at all obvious which are the most significant in body-powered prostheses' popularity. It is necessary, however, to choose one as a starting point and to devise a test for its significance. It is the opinion of the author that investigation of the aspects of the prosthesis relating to its operation (i.e.: output impedance toggling, elbow angle perception, and elbow torque perception) are likely to yield the most interesting and widely applicable insights, and so one of these will be chosen for investigation. The factors of weight, inertia, and cost are not assumed to be insignificant—on the contrary, it is recognized that they may actually be very significant factors in body-powered prostheses' popularity. However, in the larger scope of studying physical interaction and tool usage, it would be more useful to learn of their significance by elimination of the other factors, rather than by direct investigation.

Before one of the three remaining factors is selected, a brief discussion of physical interaction would be appropriate.

2.3 Physical Interaction Through Powered Tools

It is possible to model physical interaction in terms of "interaction ports" [6]—locations where elements of a system exchange energy. For example, a carpenter swinging a hammer exchanges energy with a hammer through the point where his hand meets the hammer's handle.

The physical interaction in this example can be modeled both as the carpenter providing the force and the hammer determining the velocity (the hammer is modeled as an admittance) and as the carpenter setting the velocity and the hammer determining the resulting force (the hammer is modeled as an impedance). When a tool is modeled as an impedance, then force can be viewed as a quantity that is "fed back" to the user. It will be a function of the velocity of the movement executed by the user. A more rigorous and complete discussion of the concepts of impedance, admittance, and interaction is found in[4].

In powered tools, it is possible to vary the apparent impedance of the tool (within certain limits). One example of this is the steering mechanism in a car. With non-powered steering, the inherent inertia, friction, and compliance of the system determines the torque necessary to turn the steering wheel. However, with power-assisted steering, the apparent impedance of the steering system drops considerably, reducing the torque required to operate it.

The variation in apparent impedance of a tool can also be viewed as a variation in the level of power-assist in the tool. Returning to the example of powered steering, two drivers racing through identical courses will feel different levels of force through their steering wheels, depending on the level of power-assist in each car's steering system. A car with high power-assist has a low apparent impedance, while a car with no power-assist has a high apparent impedance.

Understanding human interaction with powered tools would help engineers to optimize the tool's design to maximize performance and ease of use. For example, understanding human use of powered tools would allow engineers to design a power steering system that would provide just the right level of power-assist. The ideal steering system would have an apparent impedance high enough to allow the driver to sense road conditions (rough or icy spots) while not having an apparent impedance so high that the car requires excessive effort to operate.

An investigation into how the level of power assist affects prosthesis performance may yield insights that would be applicable to a variety of human interactions with powered tools.

2.3.1 Effects of Power-Assist on a Body-Powered Prosthesis

Now, returning to the discussion of which aspects of a body-powered prosthesis may make it functionally superior to a motorized prosthesis, it is appropriate to reconsider which of the three (output impedance toggling, elbow angle perception, and elbow torque perception) will be singled out first for investigation. It is interesting to note that that, just as power-assist in an automobile decreases the amount of steering torque perceived by the driver without affecting the relation between the position of the steering wheel and the steering angle, it is also conceivable that the addition of a power-assist to a body-powered prosthesis would change the level of elbow torque perceived by the wearer without changing the perception of the elbow angle. Given this, it appears that investigation of the significance of the level of elbow torque perceived by the wearer of a body-powered prosthesis may yield insights that would have widespread application in the development of other powered tools.

There are, of course, significant motivations for considering each of the other factors. The ability of the prosthesis to toggle output impedance levels is obviously very useful to amputees. When this capability is removed, amputees have difficulty performing tasks that otherwise pose no particular challenge.[3] Also, the wearer's perception of the prosthesis' elbow angle may well be a very significant factor, especially considering humans' tendency to preserve the kinematics of movement, even in the presence of changing dynamics.[12] However, the level of elbow torque perceived by the wearer will be initially pursued as a possible significant factor in body-powered prostheses' performance.

2.3.2 A Note on Terminology

Throughout this thesis, the phrase "the level of power-assist" is used several times. As it is used in this thesis, this phrase could technically be replaced by the phrase "varying the output impedance". In fact, this substitution might help minimize misunderstandings that come from modeling physical interaction as signal transfer. However, the author has chosen not to make this substitution in order to maintain clarity on another point.

In a powered tool, apparent impedance of the environment as perceived by the tool user can either be changed by changing the level of power-assist in the system (as in the power steering example discussed) or by changing the physical travel required to operate the tool. A simple example of the latter is a lever. For example, using an extension bar on a wrench to break loose a rusted nut changes the apparent impedance of the nut without changing the power transmitted by the person operating the wrench. Although the contact force at the handle is decreased, the travel required to perform the action is increased by the same proportion, keeping power supplied constant. However, in a power-assisted system, as the term is used in this thesis, the force required to perform a task with a tool is decreased, but the travel or motion required does not change. In this scenario, the power supplied by the user is less than in the non-power-assisted case: it is supplemented by power from the tool.

The power-assisted prosthesis discussed in this thesis will have the ability to modulate its output impedance, but it will accomplish this by changing the level of power supplied by the prosthesis, rather than by changing the range of motion required of the user. Because of this, the author has chosen to keep the phase "power-assist" in discussions that are actually about modulating the apparent output impedance of a prosthesis (as perceived by the environment) and modulating the apparent input impedance (as perceived by the user).

2.4 Working Hypothesis

In summary, the following will be accepted as a working hypothesis and will be used to guide the rest of this work:

> Body-powered prostheses are more popular than motorized prostheses because they are functionally superior. More specifically, the fact that body-powered prostheses allow the wearer to sense elbow torque make them functionally superior.

Of all the possible contributing factors listed earlier, the level of elbow torque perceived by the user has been singled out as the factor of interest. It will eventually be tested by implementing and systematically varying a power-assist on a body-powered prosthesis.

This working hypothesis will provide a line of questioning and experimentation to follow. Even if this hypothesis proves to be false, testing it should yield insight into the operation of prostheses and of powered tools in general.

Chapter 3

Evaluating Prosthesis Performance

Although a hypothesis was discussed in the preceding chapter, the details of testing that hypothesis have not yet been specified. As stated previously, performance will be tested while systematically varying the level of power-assist on a powered body-powered prosthesis emulator. However, there is not an established method for quantifying overall prosthesis performance. Such a method will be proposed and discussed in this chapter.

3.1 Choosing a Metric for Prosthesis Performance

Because a prosthesis is used for so many different types of tasks—in fact used differently by every owner—it is not feasible to find one test that categorically rates a prosthesis' performance for all people under all conditions. It is analogous to the problem of creating a standardized test to evaluate students entering graduate school. It is impossible to find one test that can adequately assess a student's strengths, aptitudes, and weaknesses in all areas, yet some objective measure of a student's scholastic ability is required. The best that has been done so far is to test the student with a representative subset of tasks. The student's performance on that subset is assumed to be correlated with his or her overall academic ability. It is a system that is far from perfect, but also one that is very widely used. For lack of a better way, a similar approach will be used here. A task that is a representative subset of many tasks performed using prostheses will be used in evaluating prosthesis performance.

The selected measure of a prosthesis' performance should be reflected by the ease of use. Specifics such as time to complete a task, or error in tracking cannot portray that. Another measure is needed.

Throughout the rest of this work, an examination how rapidly adaptation occurs is used to characterize ease of use in a prosthesis. The basis for using adaptation patterns to characterize ease of use is presented below. This metric allows a quantitative measure to be made. The measure of ease of use will be independent of subject performance on any given trial. It will only depend on the change of performance over time when the subject is exposed to a sudden change in one aspect of an otherwise familiar task. This sudden change will often be referred to as a "perturbation" throughout the rest this thesis.

3.1.1 Basis for Using Adaptation Behavior as a Measure of Performance

There is evidence that kinematics is the chief goal in many human movements. When asked to move between two points, subjects tend to use a nearly straight path, even when a path is not specified.[5] In addition, when disturbed by coriolis forces, subjects' movement paths deviate initially, but gradually converge back to a straight line as the subject adapts.[7] Even when an unfamiliar force field is imposed by a robot, subjects' trajectories show significant initial deviation, but with practice become straight again. As an internal model is formed and the forces necessary to produce a given set of kinematics are learned, adaptation occurs. The rate at which an internal model can be formed is related to the rate at which adaptation occurs.

In extending the discussion to tool usage, it is plausible that the rate at which a tool allows the user to form an internal model of the environment is related to the rate at which the user will be able to adapt to changes in the environment while using the tool. Therefore measuring adaptation patterns may provide an indirect measurement of how quickly the amputee is able to learn to reach a given kinematic objective. In other words, if an amputee can readily adapt to a change in the task being performed, the prosthesis is "easy to use".

It is important to note that this metric is only a valid measure of performance within a single subject. Due to variations in age, strength, previous experience, and skill level, different subjects may have very different adaptation patterns, even given the same tool and task. Because of this, the only meaningful measure of prosthesis performance is a *relative* one—some function of the difference in adaptation patterns when an amputee uses each of two prostheses in an otherwise identical task. This raises an important question about how much of the learning that occurs while using one prosthesis is transferred to use of the other. Although this is a significant question that is worthy of careful examination, it will not be addressed here, as it is not crucial to the remainder of this thesis.

3.2 Finding a Task in Which Measurable Adaptation Occurs

Before adaptation patterns can be compared, they must first be obtained. In order to obtain adaptation patterns it is necessary to identify a task that can be used as a context.

In order to be a candidate for inducing adaptation in amputees, a task should involve some contact with a kinematic constraint. The requirement that the task involve a kinematic constraint increases the chances that performance will depend on the level of power assist. In free motion tasks there is no direct contact with the environment. In constrained motion tasks, however, the amputee is required to form some kind of mental model of the environment with which the prosthesis is interacting. Although, this stacks the cards in favor of a positive result (i.e.: finding that the level of power assist is a significant factor), it would also make a negative result that much stronger.

In addition, constrained motions have not been widely studied. This also is motivation for pursuing a constrained motion task.

It is also necessary that the task not require high elbow extension torques. All elbow extension in body-powered prostheses is actuated by gravity, rather than by the wearer. Requiring high elbow extension torques may make it impossible for the amputee to execute the task easily.

3.2.1 Crank Turning as a Candidate Task

The candidate task chosen to be pursued in this work is crank turning while seated. More specifically, subjects will be asked to track a random target signal with the crank, while seated. In addition to meeting all the requirements listed above, crank turning has been studied by several different researchers in the Newman Laboratory already, both with amputee and with non-amputee subjects. This precedent gives an experience base upon which to draw and a benchmark against which to compare the experimental results.

3.3 Finding an Appropriate Perturbation

It is also necessary to select an appropriate way to change or perturb the task in order to induce adaptation.

In order for a change to the task to be a candidate perturbation, it must be difficult enough to require some time for adaptation. If the adaptation occurs too quickly, it will be impossible to measure it. At the same time, the task must not be so difficult that the subject does not show adaptation within the time available in the laboratory.

In addition, the perturbation must change the force required to execute the task. This requirement is another way to stack the cards in favor of finding elbow torque information to be a significant factor in prosthesis performance. Again, if elbow torque information is found *not* to be a significant factor, then having this type of perturbation would make that only a stronger result.

3.3.1 Position-Dependent Crank Torque as a Candidate Perturbation

The perturbation that has been chosen for this experiment is the addition of a position-dependent torque about the axis of the crank. It will be implemented by hanging a weight from a cable that is wrapped around a non-circular lobe that it attached to the crank shaft. The moment arm of the weight about the axis of the crank will depend on the crank's position and on the shape of the lobe. There is also an acceleration-dependent torque due to the hanging weight. The specific design of this lobe will be discussed in more detail in the next chapter.

In summary, the adaptation pattern of an amputee using a prosthesis has been proposed as a measure of prosthesis "ease of use". A crank turning task perturbed by a position-dependent torque has been proposed as a task in which adaptation may occur. The following chapter details the development of an experiment to determine whether the proposed task and perturbation actually do induce adaptation.

Chapter 4

Hardware

This chapter contains a description of the apparatus used to conduct the experiment. It includes descriptions of the power-assisted prosthesis emulator, the crank, the computer used to supervise the experiment, and safety precautions taken.

4.1 Overview

The experiment described in this chapter has to date been performed by one non-amputee subject and one amputee subject. Some parts of the description (namely those that concern the prosthesis emulator) apply only to the amputee subject.

The main pieces of hardware used during the experiment are the prosthesis emulator, the crank, instrumentation electronics, and a desktop computer. While the subject sat in a chair and turned the crank with the prosthesis emulator (or with an unimpaired limb) the computer generated a target signal, recorded data from the experiment, and supervised the experiment to ensure that velocity limits on the emulator were not exceeded.

4.2 Prosthesis Emulator

The prosthesis emulator referred to in this experiment was built in 1981 by Cary Abul-Haj as part of his Masters Thesis. It is a piece of experimental hardware that was built in order to simulate a number of different prostheses on a single, instrumented platform. A detailed description of its design and construction can be found in the thesis[1] and will not be repeated here.

The only modification to the emulator's functionality since its creation has been the addition of a cable that allows it to simulate a body-powered prosthesis. The cable allows the wearer to move the emulator even when it is unpowered.

4.2.1 The Power-Assisted Emulator

In order to allow the emulator to have power-assist capability, a feedback loop was implemented. A uniaxial force transducer was placed in series with the cable such that all tension in the cable was measured by the force transducer. This tension measurement signal was then sent to a variable gain amplifier. The output of this amplifier was then sent as the control input to a 400W current-controlled amplifier. This amplifier powered the DC direct-drive motor on the emulator. In this way, the torque produced by the emulator's motor was proportional to the tension in the cable.

It is interesting to note that even when the prosthesis is unpowered, the nature of the mechanical system is such that the net torque about the elbow is roughly proportional to the tension in the cable. All the power-assist system does is to give the user a "boost" in performing whatever action was intended. See the section titled "Physical Interaction Through Powered Tools" (section 2.3) for a more thorough description of the effects of the power-assist on the emulator's dynamic behavior.

It is important to note that the kinematic relation between the amputee's body posture and the position of the prosthesis remains unchanged as the level of power-assist is varied. The power-assist does not change the *motion* required to execute a task, only the level force required.

4.2.2 Stability

The prosthesis emulator is a powered system, and so may be capable of instability. It is not difficult to imagine a scenario where instability could occur, especially with very high levels of power-assist. With a small initial pull on the cable, the emulator could flex powerfully and generate enough torque to not only lift the forearm, but to pull the socket and shoulder of the amputee forward. This is a motion in turn pulls on the cable harder, which increases the torque at the elbow. The result would be an instability that could potentially be very dangerous.

Although dynamic models of the emulator/amputee system proved to be too unwieldy for a rigorous stability analysis to be practical, qualitative observation of the emulator's operation in making unconstrained motions revealed no tendency toward pathological instability, even when the power-assist was very high. However, during the experiment, the power-assist was always kept at half of the highest tested level in order to ensure a comfortably large margin of safety.

4.3 The Crank

The crank used during the experiment was also built by Abul-Haj and is described in his Ph.D. Thesis. It is a metal crank, mounted on the edge of a heavy-duty table. Its axis of rotation is horizontal to the floor, so in order to turn it, one moves the handle in a circle in a vertical plane. A more complete description of the crank and its instrumentation is included in Abul-Haj's thesis[2].

Of the instrumentation described by Abul-Haj, only the crank angle potentiometer remains. Crank handle contact forces are now measured by a six-axis force transducer manufactured by ATI, although that sensor was not utilized in this experiment.

The crank's handle is located 22 centimeters from the axis of rotation. The counterweight was adjusted so that, with the handle in place, the crank's center of gravity lay nearly on the axis of rotation.

4.3.1 Calibration

The crank angle potentiometer was calibrated by interpolating between measurements taken at top dead center and bottom dead center and found to be

$$\theta = -38.96 * (V - 9.80),$$

where V is the voltage read by the computer's analog-to-digital card, and θ is the angle of the crank in degrees with zero degrees being bottom dead center and with 90 degrees pointing directly toward a right-handed crank turner.

4.3.2 The Lobe

A long rectangular block, 38 millimeters by 117 millimeters was attached to the shaft and a cable was wrapped around it. From the cable it was possible to hang weights as desired. The moment arm of the weight changed from 19 to 75 millimeters–a change of nearly 400%–within 76 degrees of rotation. See figures 5-1 and 5-2 for a graphical description.

4.4 Desktop Computer

Data was collected by and stored on a 100 MHz Pentium PC operating in QNX, a real-time variation of the UNIX operating system. The computer's analog-to-digital card read signal voltages as they came from the instrumentation electronics. The software sampled the signals 1000 times per second and recorded the data in a separate file for each experimental trial.

4.5 Safety Precautions

The computer program that supervised the experiment also monitored the speed of the emulator's elbow flexion. If the elbow velocity exceeded a certain pre-set limit (about six radians per second), the program shut down power to the emulator. In addition, both the subject and the experimenter had access to "kill" switches that could disconnect the emulator's power supply. Fortunately, the emulator showed no signs of unstable or otherwise undesirable behavior, and none of these safety measures were required during the course of the experiment.

Chapter 5

Experimental Procedure

This chapter contains a description of the procedure followed during the experiment. It includes a description of a typical experimental trial, the sequence of experimental trials executed in the experiment, and the data recorded during each trial.

5.1 Setup

Before beginning the experiment, each subject was given a copy of the Informed Consent Document (see Appendix D), and was asked to thoroughly read it. The author answered the subjects questions, if there were any, and the subject signed the form.

Preparing the amputee subject for the experiment required that the emulator be attached to the amputee's socket and that the author assist the amputee in donning and learning to operate the emulator.

5.1.1 Seating

Each subject was seated in a flat steel–backed chair so that the crank was within easy reach and the range of motion required was achievable. At eye level, approximately one meter in front of each subject was a computer display. The non-amputee subject was seated such that the crank turned in the subject's saggital plane.

The amputee subject had a limited range of motion compared to the non-amputee subject, and reorientation of the amputee's chair was necessary to reduce the range of motion required for the task. The amputee subject's chair was pulled around in front of the crank apparatus such that the subject's saggital plane made a 53 degree angle with the crank's plane of rotation. In any future experiments, the amputee subject will be seated with the crank's plane of rotation in the saggital plane, but the crank length will likely be shortened in order to reduce the required range of motion.

During the experiment, a strap ran from the top of the chair back, over the non-amputee subject's right shoulder, to the bottom of the chair back. This prevented excessive shoulder movement during cranking. The amputee subject was restrained by a lap belt as well as by a strap that ran from the top of the chair back, over the amputee's right shoulder, across the front of the amputee's chest, and to the bottom of the chair back at the amputee's left side. The amputee was restrained differently in order to allow enough freedom of motion to operate the prosthesis.

Each subject received an explanation what they would experience during the experiment and were reminded of the goal of the experimental trials—to try to track the target signal as closely as possible.

In general, the amputee subject could not comfortably exert forces as high as could the non-amputee subject. Because of this, the experimental setup varied between the subjects. For the non-amputee subject, the weight had a minimum moment arm at a crank angle of near 45 degrees. For the amputee subject it was near 90 degrees. The lobe was placed differently for the amputee subject because it decreased the average

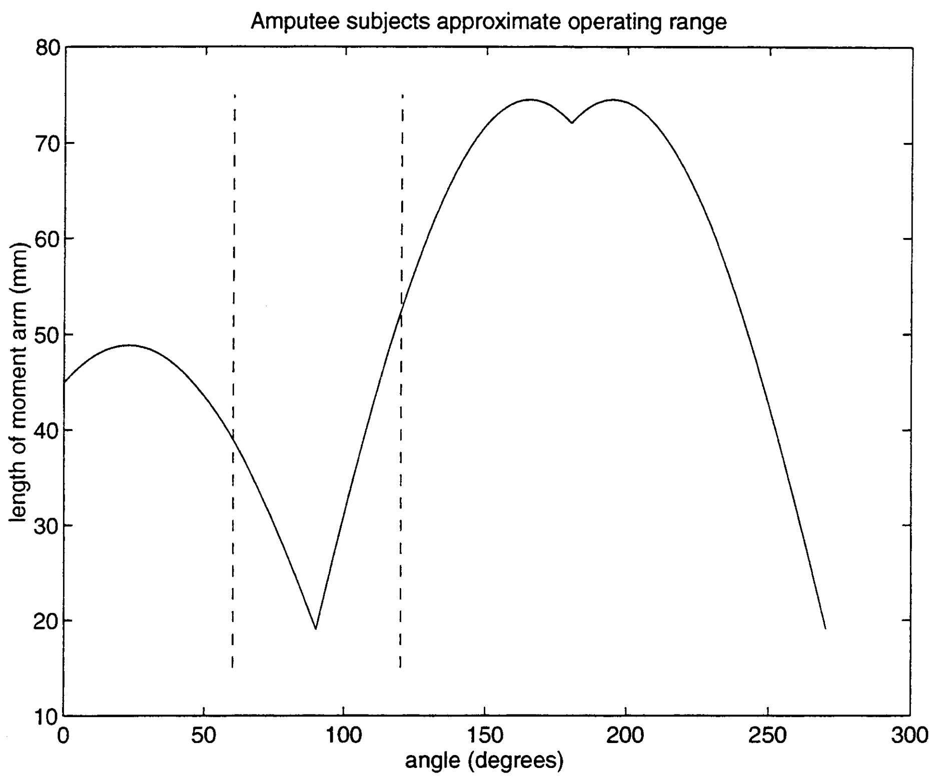


Figure 5-1: The moment arm to angle relationship of the lobe during amputee trials. The vertical lines show the approximate range of the amputee subject's motions.

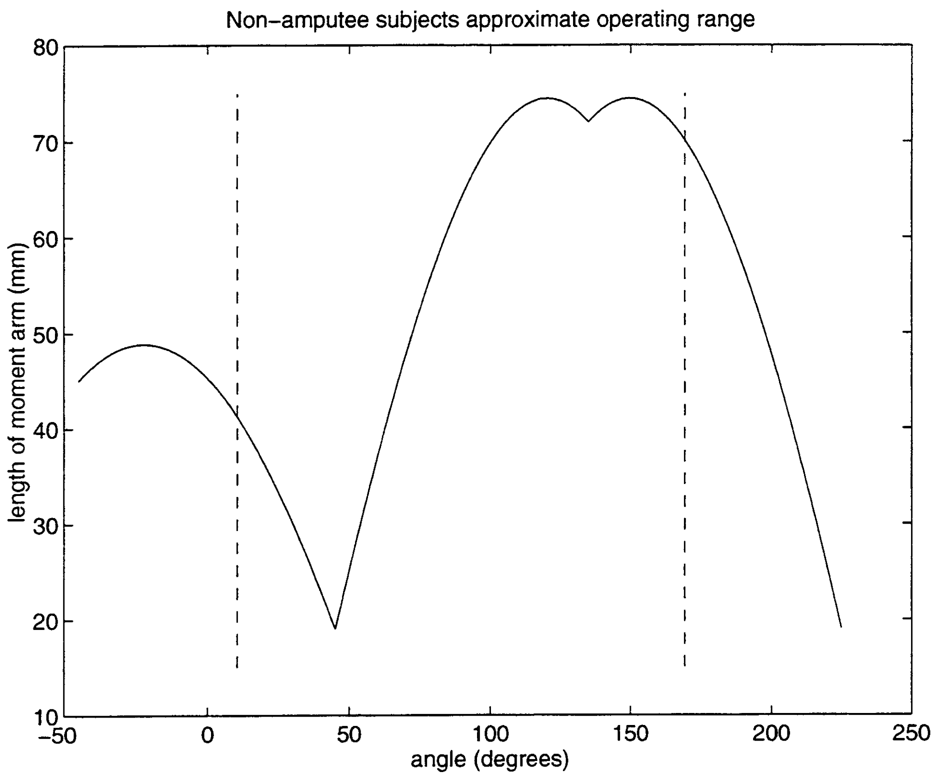


Figure 5-2: The moment arm to angle relationship of the lobe during amputee trials. The vertical lines show the approximate range of the non-amputee subject's motions.

torque required to turn the crank, while maintaining the large variation of torque with position. See figures 5-1 and 5-2 for a graphical description of the torque-angle relationship.

The non-amputee subject performed the experiment with 9.0 kilograms of mass hanging from the cable, while the amputee had 2.3 kilograms. Again, the differences were put in place in order to minimize the forces required of the amputee subject.

Target Signal

The computer also generated a random target signal for the subject to track during each trial, and displayed the signal, as well as a representation of the crank's position, on a computer monitor located in front of the subject. The signal consisted of a sum of six sine waves (multiples of $\frac{1}{6}$Hz up to 1 Hz) of random amplitude and phase[1]. This signal was displayed on the computer screen in front of the subject during each trial.

5.2 Trials

At the beginning of each trial, the computer screen in front of the subject was white and blank. After about one second, two points appeared near the left side of the screen—a heavy point (approximately 3 mm in diameter) and a smaller point (approximately 1 mm in diameter)—and began to move slowly across the screen. The two points traced a path across the screen and, six seconds later they both winked out near the right hand side of the screen.

The small point represented the target signal, and the heavy point represented the

[1]To be more specific, the signal was composed of six sine waves and six cosine waves, one of each at multiples of $\frac{1}{6}$Hz. Each wave was assigned a random amplitude (produced by the computer's internal random number generator), but no phase lag. However, summing a random amplitude sine and a random amplitude cosine wave of the same frequency is essentially equivalent to producing a sine (or cosine) wave of random amplitude and phase.

position of the crank. The display was scaled such that zero degrees (bottom dead center) was at the bottom of the screen and 180 degrees (top dead center) was at the top. During each trial the subject made an effort to track the randomly varying target, such that the heavy point followed the small point exactly. There was roughly a 5 second pause between each trial while the data was saved.

The subject was given the opportunity to rest briefly (about 30 seconds) after each 20 trials.

After 80 trials, the subject was asked to stand up and take a longer break. The amputee subject removed the emulator, and both subjects walked around for about five minutes. Then they each sat back in the chair for the next block of 80 trials.

During the next block of trials, a weight was added (9.0 kg for the intact subject, 2.3 kg for the amputee subject) to the cable that hung from the lobe. This constituted a perturbation to the task. Each subject then completed an otherwise identical 80 trials.

After a second 5-minute break, each subject completed 40 additional trials with no weight hanging from the lobe.

5.3 Data

During each trial, the following data were sampled at 1 kHz and recorded:

1. cable tension
2. input signal to power amplifier (nominally proportional to motor current)
3. emulator elbow position
4. emulator elbow velocity
5. emulator elbow torque

6. crank angle in volts

In addition, the crank angle in degrees was calculated, as was the error (i.e.: the difference between the crank angle and the target angle). All of the above quantities were stored in a data file after each trial. These files were stored without text headers so that they are very easy for MATLAB (R) to load and manipulate.

The following chapter contains the results obtained from the experimental trials as well as some analysis indicating the extent to which adaptation was observed.

Chapter 6

Results and Analysis

This chapter presents the results of the experiment and the accompanying analysis. It includes plots of each subject's performance and an attempt to describe the perturbation data with three different adaptation models.

6.1 Data Presentation

6.1.1 Crank Angle and Target Angle Data

The samples of data shown in figures 6-1 and 6-2 show the crank and and target angle data for two typical trials and are representative of their respective sets of trials. The non-amputee target signals had, on average, larger ranges of motion by a factor of two. Both subjects seemed to have no problem tracking the general shape of the target after a the first few trials (as shown in figures 6-1 and 6-2).

The downward spike in both graphs at 5000 ms is also typical of the graphs. Some undetermined artifact of the code or hardware made an identical 6 ms spike in every trial's data. This spike was removed by interpolating between the data points on either side of it.

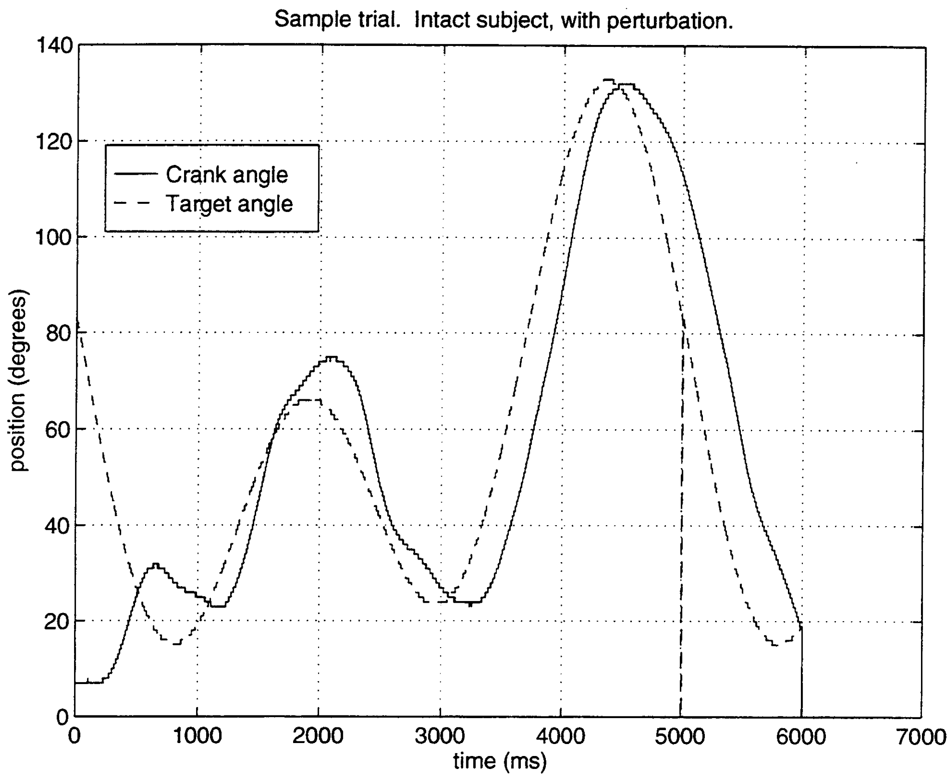


Figure 6-1: Data from a typical non-amputee trial

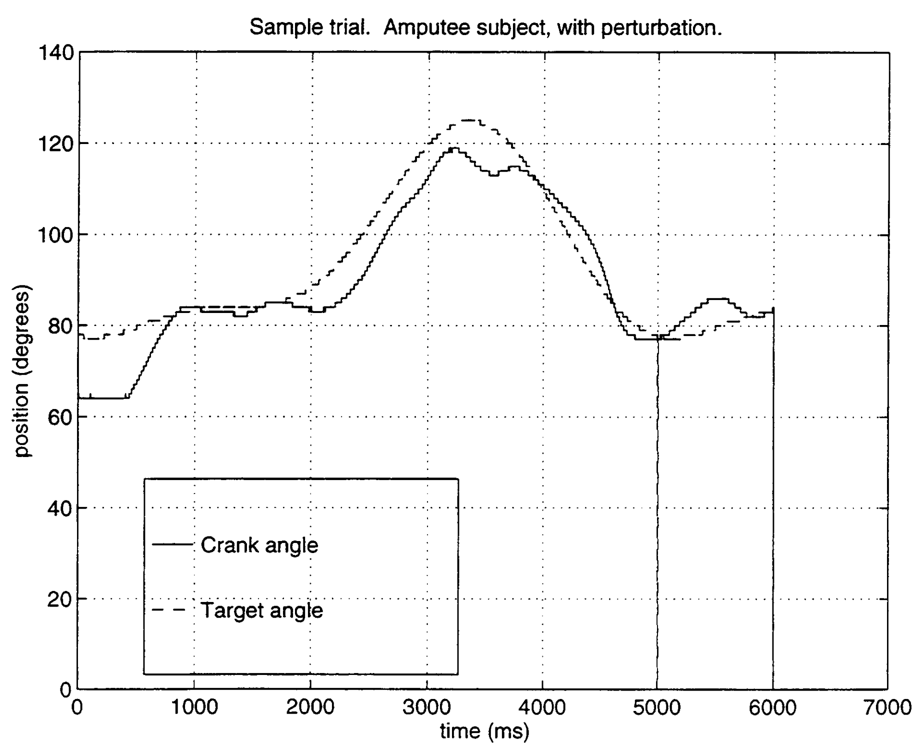


Figure 6-2: Data from a typical amputee trial

The second downward spike at the end of each data set is an artifact of how the data were recorded. That data point does not enter in to any of the analysis.

6.1.2 Data Processing

The data were gathered at 1 kHz over 6 seconds, resulting in 6000 data points. (Actually there were 6001, but the final point was always zero and is always ignored.) The column of data that recorded the distance between the crank angle and the target angle–or the error–was stripped out of each data file and fed into a MATLAB (R) data processing script.

The first second of data was ignored. That ensured that the subject had some time to pick up the target signal and home in on it before the error started being counted. That left 5000 points of error data.

The mean square of the 5000 data points was calculated in order to reduce the subject's total error over the trial to a single scalar measure of the error. This removed any information about when during the trial the error occurred and whether the error followed any particular pattern over the course of the trial. However, for this analysis, that information will not be needed and so can be safely set aside.

In order to partially compensate for differences between target signals, the mean square error was divided by the variance in the target signal for a given trial. As can be seen from Appendices B and C, the amplitude of the target signal varied significantly between trials. Normalizing the mean square error in this way compensates to some degree for the variation in difficulty between trials.

It should be noted that due to the differences in the task between the non-amputee and the amputee subject that it is not necessarily meaningful to compare even their normalized mean square error values. The difference in lobe position, weights used, amplitude of crank motion required, and seating orientation change the apparatus sufficiently that it no longer constitutes the same task. IT should also be noted that the data files for the non-amputee's last five trials during the perturbation were not stored properly by the computer, and so data exists for only the first 75 of those trials.

6.1.3 Normalized Mean Square Error Plots

The normalized mean square error for each trial was calculated and then plotted. What follows in figures 6-3 through 6-8 are the normalized mean square error plots for the trials before, while, and after the perturbation was introduced for both the non-amputee and amputee subjects.

6.2 Interpreting the Data

6.2.1 Finding a Performance Level Prior to Perturbation

Inspection of the data shows a large amount of variability between consecutive trials. Qualitatively, no obvious steady-state level presents itself. In order to find a pre-perturbation 'steady-state' performance level, it is necessary to artificially infer one from the data. Here an average of the 41st through 80th trials will represent pre-perturbation 'steady-state'. These levels are tallied up in table 6.1.

6.2.2 Finding a Performance Level After Perturbation

Again, variability masks whatever steady-state level may underlie the data. An average of all 40 post-perturbation trials serves here as a 'steady-state' performance level. Table 6.1 shows these values.

The relative increase in the non-amputee subject mean average square error after the perturbation was removed suggests the subject may have experienced fatigue or

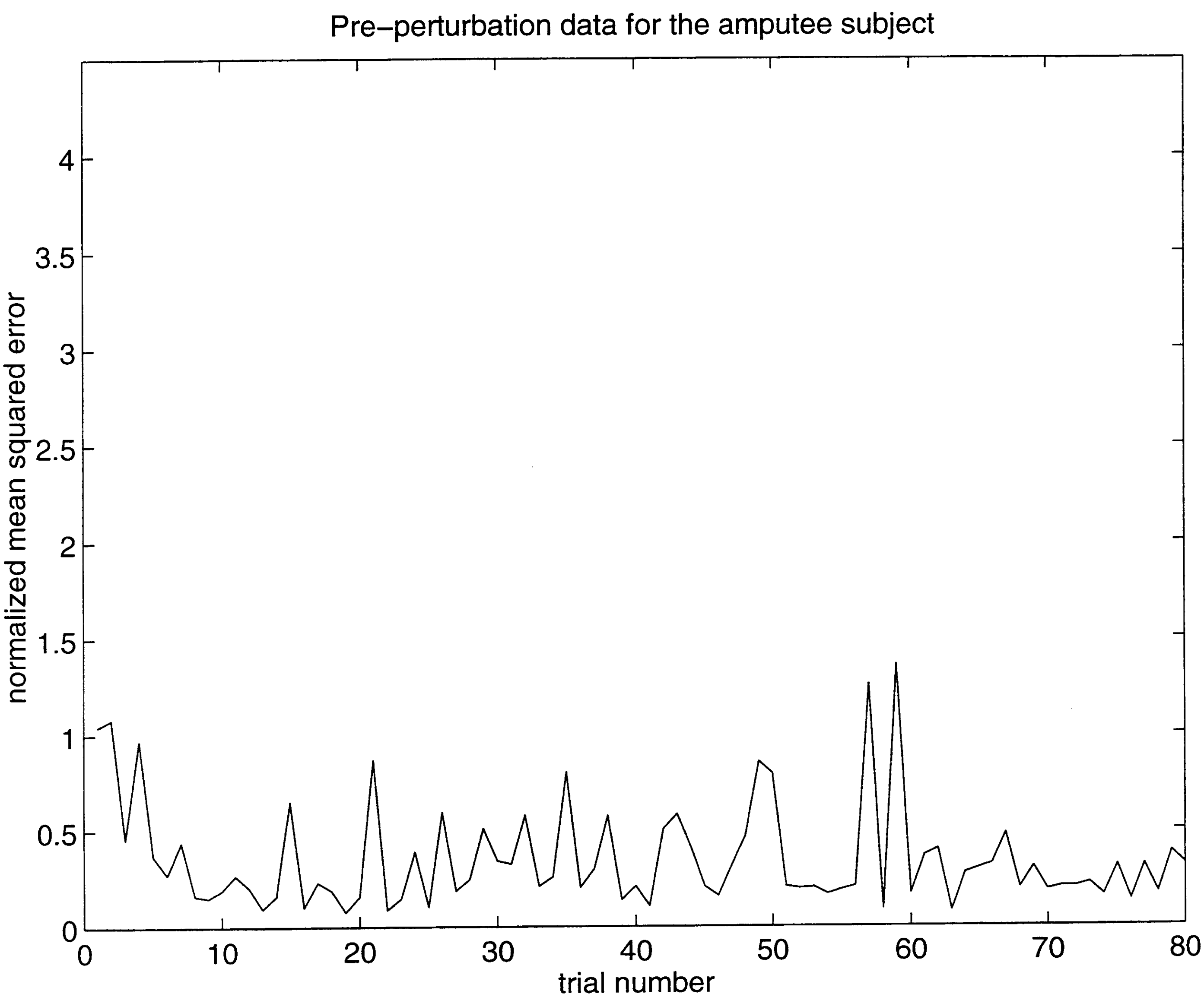


Figure 6-3: Normalized mean squared error for the amputee subject's pre-perturbation trials

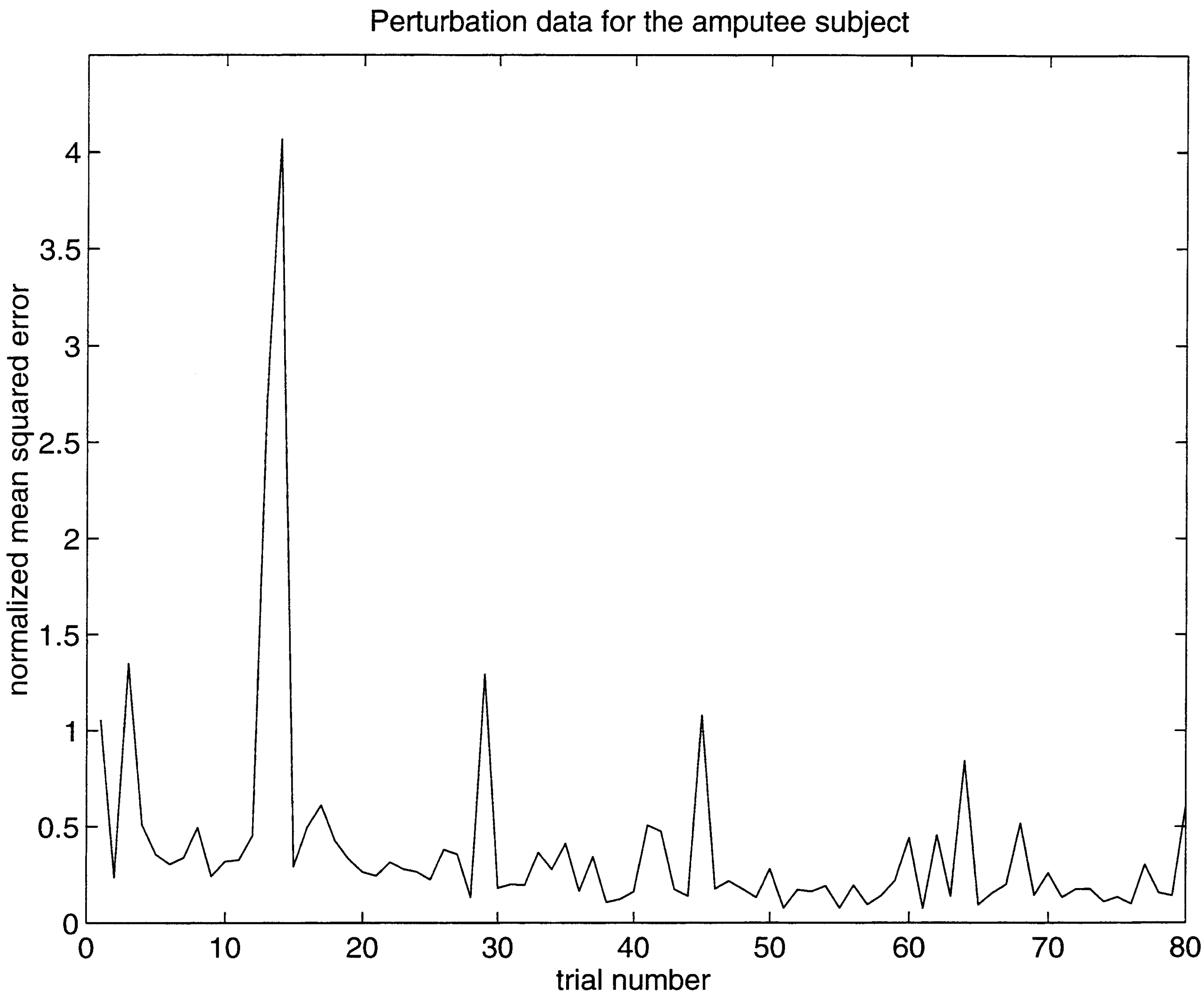


Figure 6-4: Normalized mean squared error for the amputee subject's perturbation trials

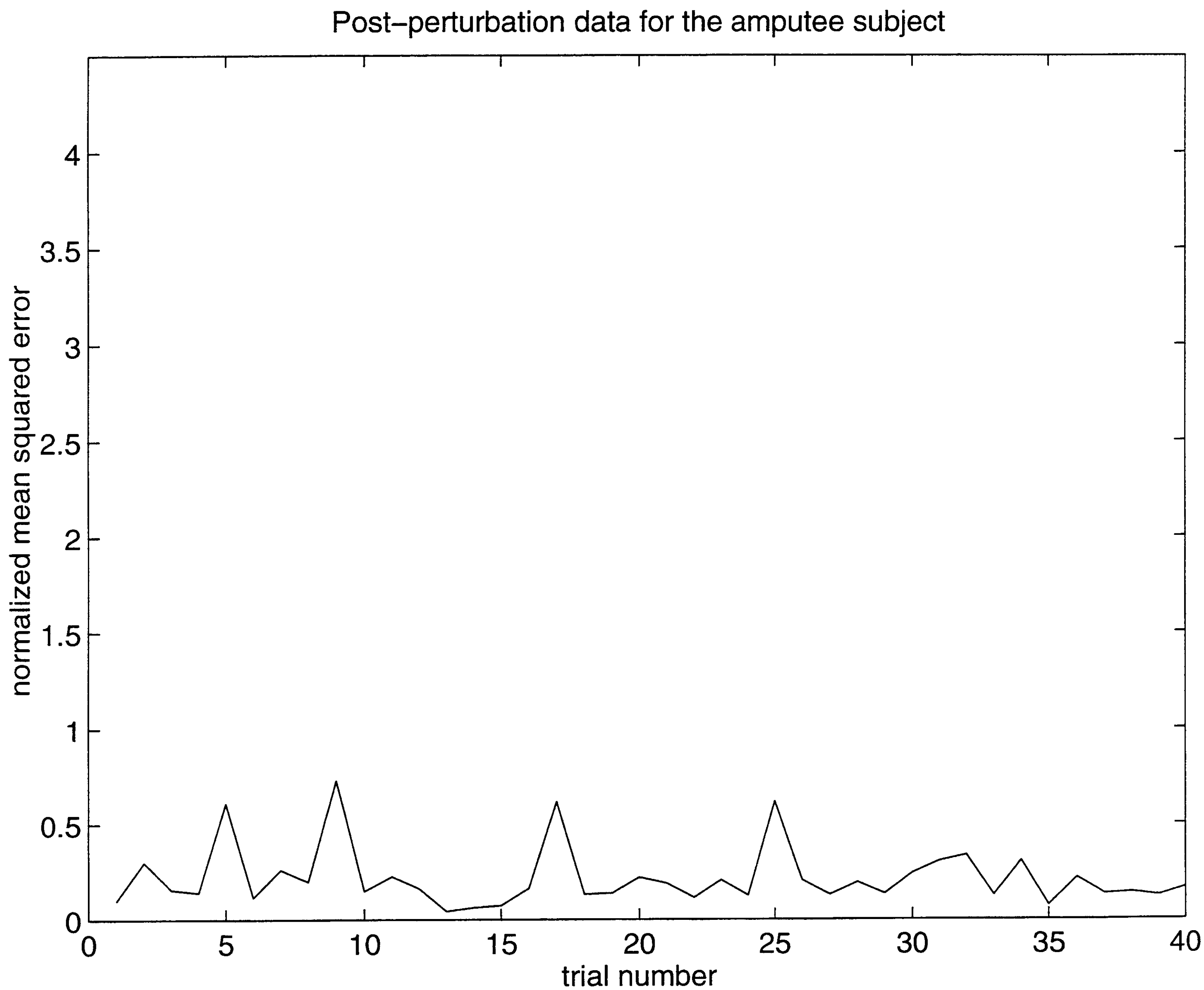


Figure 6-5: Normalized mean squared error for the amputee subject's post-perturbation trials

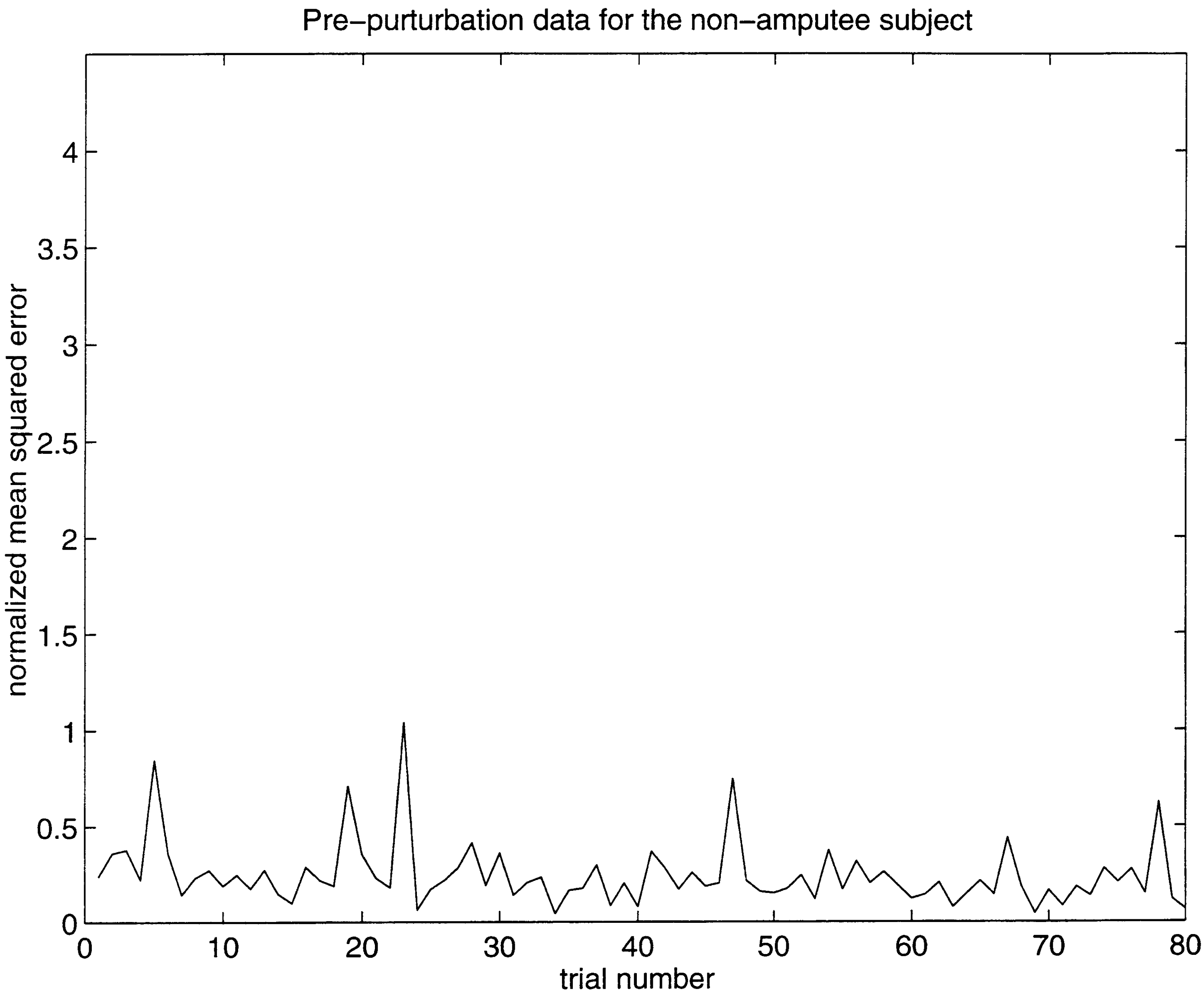


Figure 6-6: Normalized mean squared error for the non-amputee subject's pre-perturbation trials

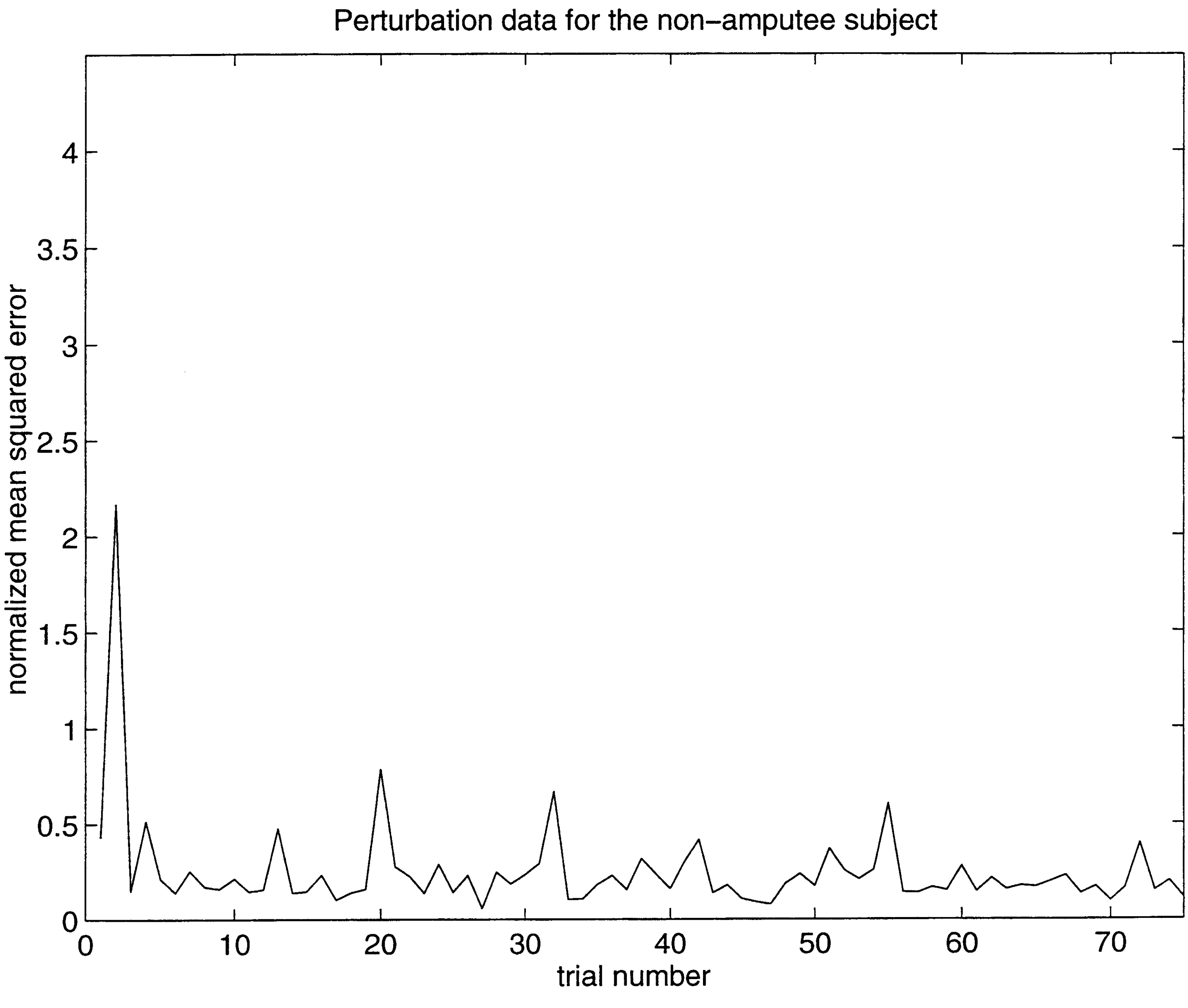


Figure 6-7: Normalized mean squared error for the non-amputee subject's perturbation trials

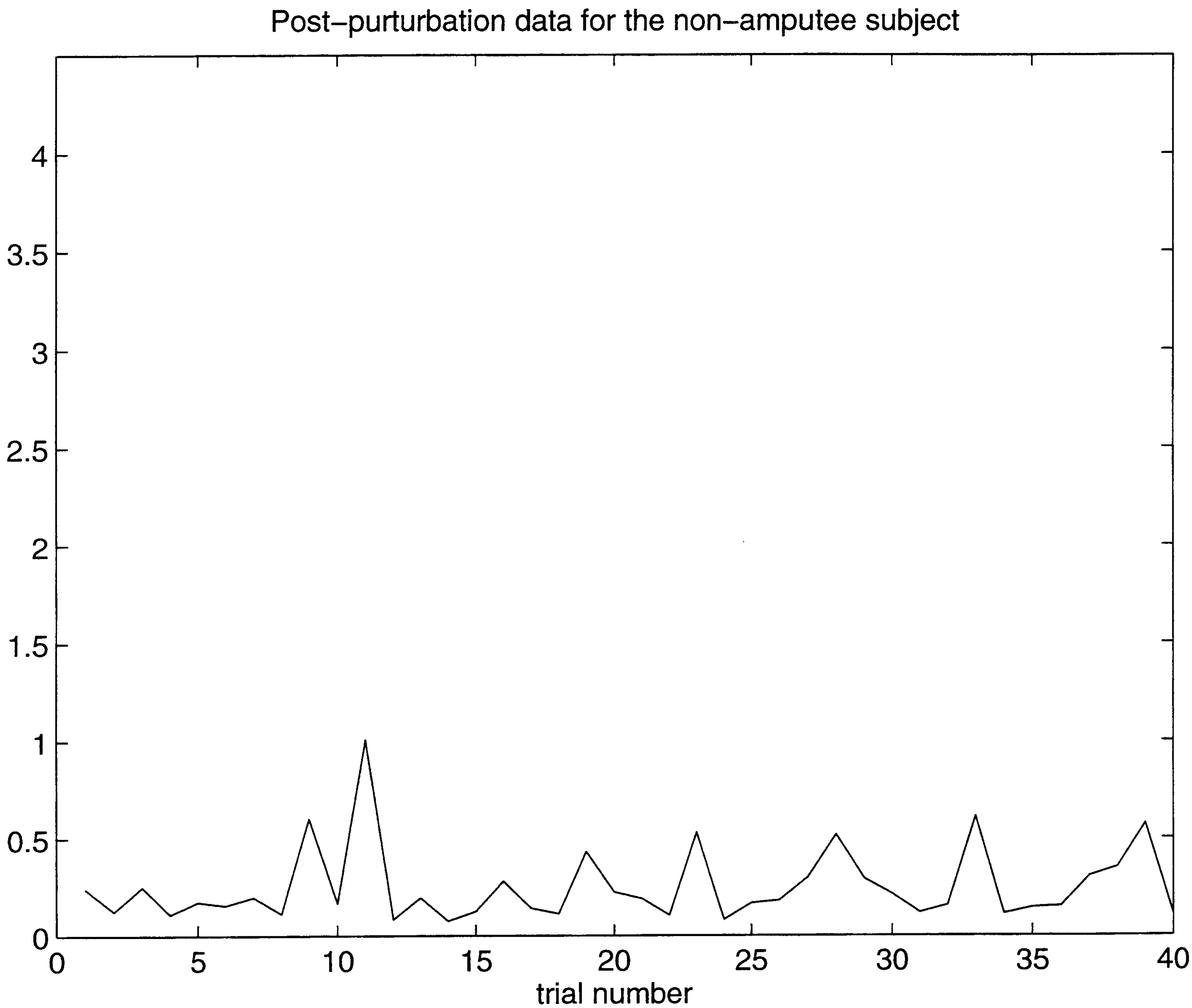


Figure 6-8: Normalized mean squared error for the non-amputee subject's post-perturbation trials

	Non-Amputee Subject	Amputee Subject
Pre-perturbation	190.3	84.2
Post-perturbation	248.6	83.5

Table 6.1: Steady-state performance levels of both subjects before before the introduction of and after the removal of the perturbation

distraction during the final 40 trials of the experiment. The subject's comments and behavior did not reflect any physical or mental fatigue, however.

The amputee subject's nearly constant level of performance on either side of the perturbation suggests that no significant fatiguing took place, nor did any significant learning of the underlying task occur during the perturbation.

6.2.3 Fitting a Model to the Perturbation Data

Due again to the large amount of seemingly random variation in the data, it is difficult to see qualitatively what, if any, adaptation pattern underlies it.

In order to make clear any underlying trends in the data, it was filtered using a sliding bin median filter. The size of the bin was determined by plotting the normalized mean square error data in a histogram and roughly fitting a chi-squared distribution to the data. The number of degrees of freedom that caused the chi-squared distribution to appear to give the best fit (between 10 and 15) was used as a starting point for the size of the sliding bins. The number of bins finally chosen was 15. A median filter was chosen in order to prevent the minority of very high-error trials from overshadowing the effects of the rest of the data. The resulting plots of the error in the presence of the perturbation is shown below (see figures 6-9 and 6-10).

Here three models are considered as possible fits for the performance data obtained in the presence of the perturbation.

- A flat line (A zero adaptation model)

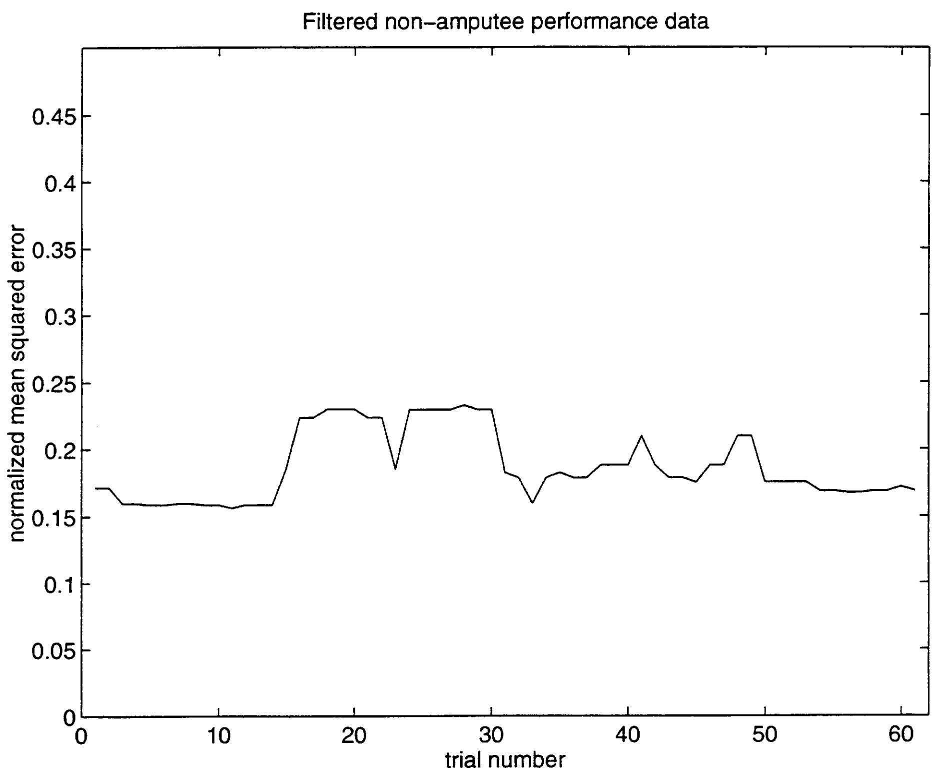


Figure 6-9: Filtered performance plot of the non-amputee subject in the presence of the perturbation. Note that implementation of the median filter with a bin size of 15 decreased the total number of data points by 14

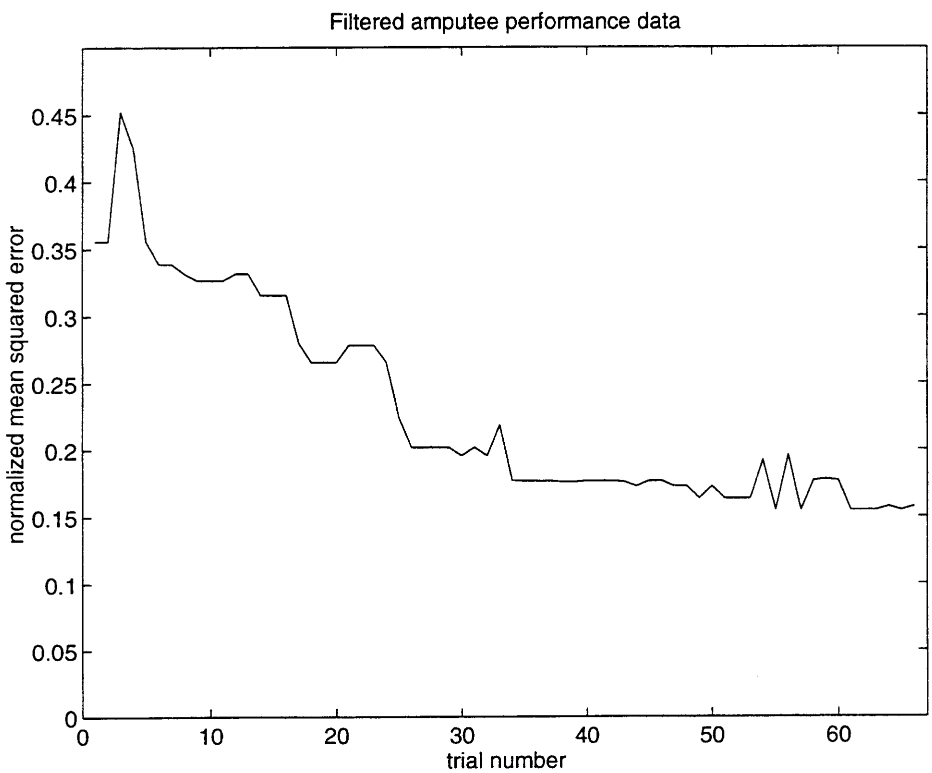


Figure 6-10: Filtered performance plot of the amputee subject in the presence of the perturbation. Note that implementation of the median filter with a bin size of 15 decreased the total number of data points by 14

- A sloped line (A constant rate adaptation model)
- An exponential curve (An asymptotic adaptation model)

A zero adaptation model might be expected to fit in the case where adaptation occurs very rapidly, and performance very quickly converges to a steady-state level. A sloped-line adaptation model would most likely fit data in which no steady-state performance level had been reached, and the subject was still adapting to the task just as actively at the end as at the beginning. An asymptotic adaptation model would probably fit best data in which measurable learning occurs near the beginning of the set, but slows toward the middle and end, suggesting that a steady-state performance level is being approached.

Plots showing each fits for the non-amputee subject follow in figures 6-11, 6-12, and 6-13 and plots showing fits for the amputee subject follow in figures 6-14, 6-15, and 6-16. Each fit was made by minimizing the rms error between the data and the model.

In order to compare the goodness of fit of each of these models, the following formula was used:

$$\frac{\sum_1^n y_{measured}^2}{\sum_1^n y_{fitted}^2}$$

where y is the normalized mean squared error and n is the number of samples. Given that the fit was obtained by minimizing the rms error (which was the case here), the goodness of fit measure will vary between zero and one with one being a perfect fit. These values are shown in table 6.2. Another (albeit non-standard) measure of goodness of fit, the rms error for each of the fits, is shown in table 6.3.

The goodness of fit measures in both subjects are the highest for the asymptotic adaptation model. For the amputee subject, the exponential adaptation model's goodness of fit meaure is 8% higher than that of the sloped line model. Of the three,

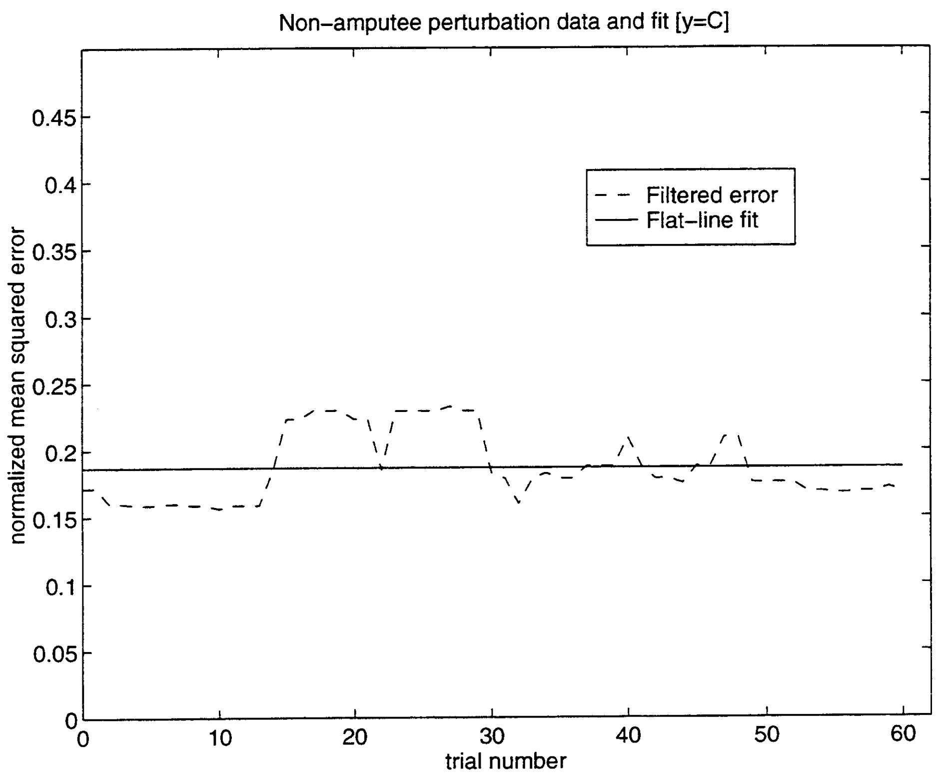


Figure 6-11: Flat line $y = C$ fit to non-amputee subject's perturbation trials. Best parameter is $C = 0.1867$

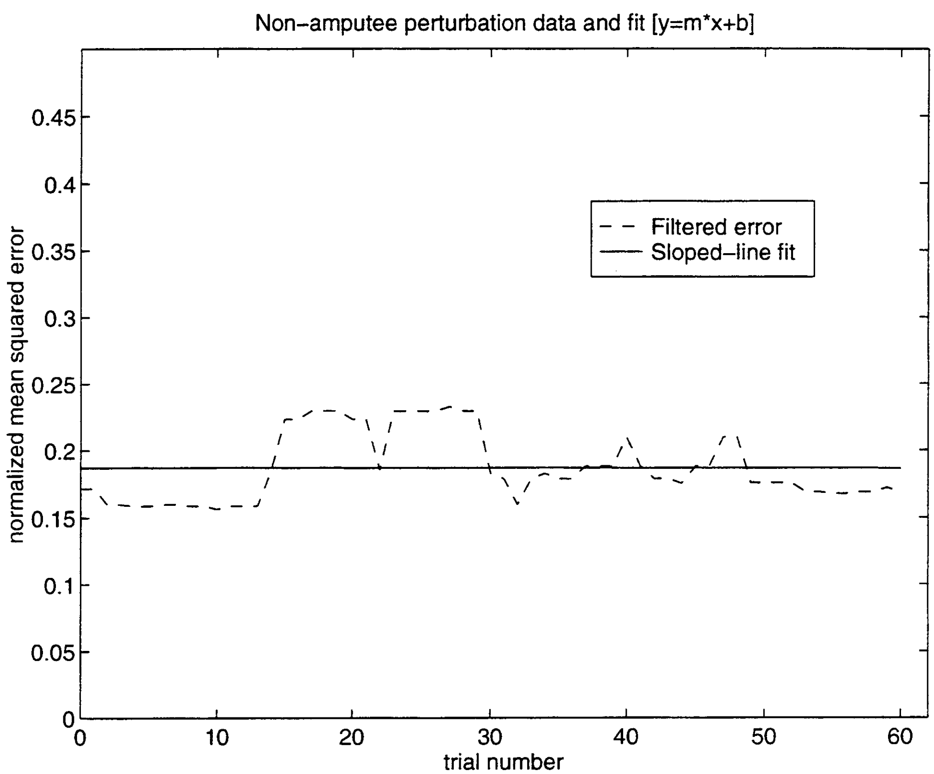


Figure 6-12: Sloped line $y = m * x + b$ fit to non-amputee subject's perturbation trials. Best fit parameters are $b = 0.1871$ and $m = -1.2046e - 05$

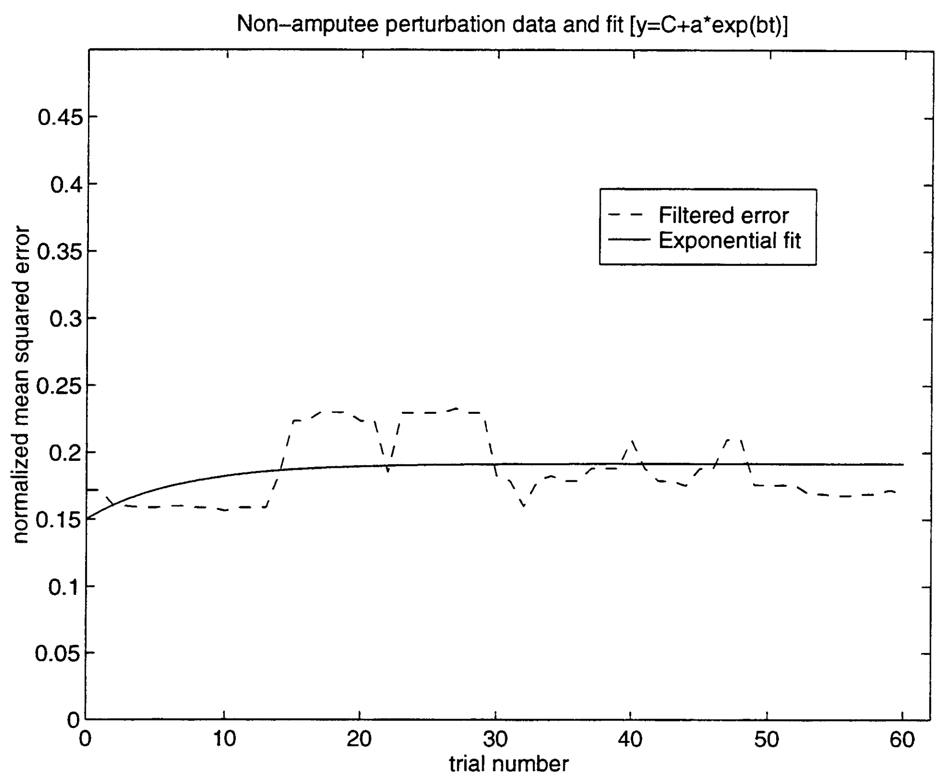


Figure 6-13: Exponential curve $y = C + a * exp(bt)$ fit to non-amputee subject's perturbation trials. Best fit parameters are $a = -0.0416$, $b = -0.1479$, and $C = 0.1917$

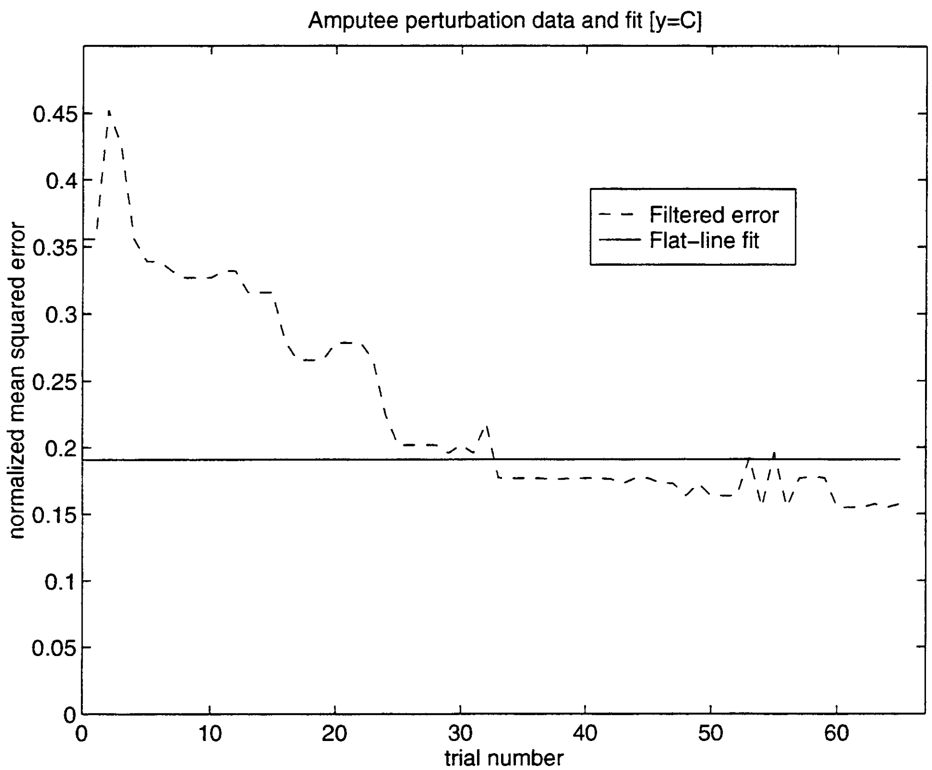


Figure 6-14: Flat line $y = C$ fit to amputee subject's perturbation trials. Best parameter is $C = 0.1910$

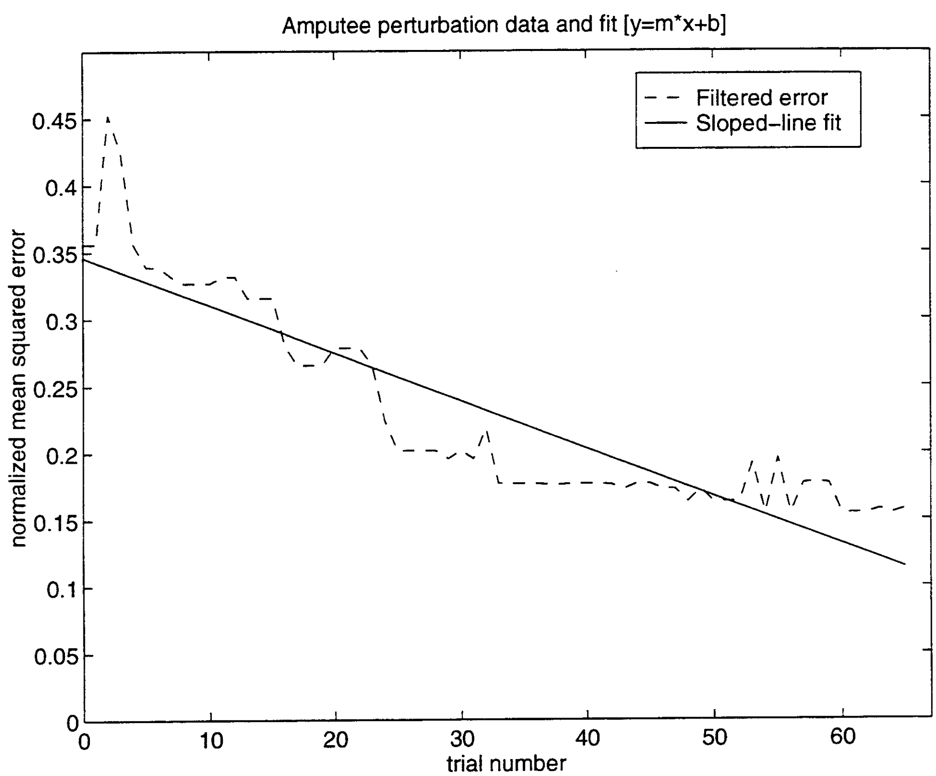


Figure 6-15: Sloped line fit to amputee subject's perturbation trials. Best fit parameters are $b = 0.3460$ and $m = -0.0036$

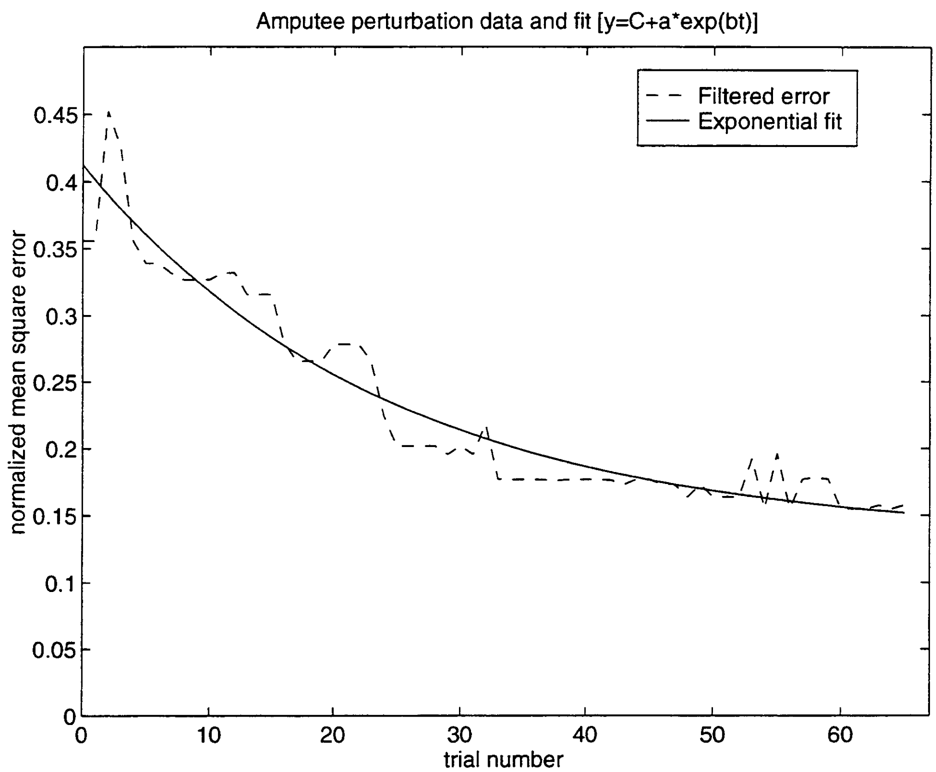


Figure 6-16: Exponential curve $y = C + a * exp(bt)$ fit to amputee subject's perturbation trials. Best fit parameters are $a = 0.2801$, $b = -0.0412$, and $C = 0.1325$

	Amputee	Non-Amputee
Flat Line	0.4187	0.9093
Sloped Line	0.8849	0.9097
Exponential Curve	0.9649	0.9203

Table 6.2: Goodness of fit for 3 models of adaptation fit to each subject's performance data. 1.00 is a perfect fit.

	Amputee	Non-Amputee
Flat Line	0.6948	0.1993
Sloped Line	0.2687	0.1993
Exponential Curve	0.1695	0.1856

Table 6.3: Root-mean-square error with 3 models of adaptation fit to each subject's performance data

the amputee subject's data is most closely reflected by the exponential adaptation model.

The non-amputee subject's data appears to be best fit by a flat line (at least to the first order). This is supported by the fact that the sloped line that best fits the data is very nearly flat. Even the fitted exponential curve is flat for the majority of the trials. Although the exponential curve fits the data slightly better (see tables 6.2 and 6.3), this may simply be due to the fact that exponential adaptation model has more parameters to vary when fitting the data. Also, the fact that the fitted exponential curve increases over time rather than decreases (suggesting performance degradation or un-learning of the task) indicates that the model may not be appropriate to the data.

It is possible that another plausible model of adaptation would fit the data better than any of the three that were applied. However, it is more likely that adaptation occurred so quickly as to not be apparent in the data, and that the observed variability is due to factors other than adaptation. In either case, the non-amputee subject's performance data appears to exhibit a zero adaptation pattern rather than the exponential adaptation pattern hypothesized earlier in this thesis.

The following chapter contains a discussion about what conclusions may be drawn from the analysis presented here.

Chapter 7

Conclusions

7.1 Implications of This Work

The comparisons in tables 6.2 and 6.3 show that an asymptotic adaptation model fits the data better than the other two proposed models. Qualitative examination of the amputee subject's performance data during the perturbation in figure 6-4 does indeed suggest a downward trend. This indicates performance improvement over time, but the details are obscured by the variability in the data. Similar examination of figure 6-7 also shows an apparent downward trend, mainly in the decrease of the maximum height of the outlying data points. It may be that it is these outliers (which were filtered out in this analysis) hold some information about adaptation.

In the amputee subject, the fact that the asymptotic adaptation model fit better than the no adaptation model suggests that adaptation occurred slowly enough to be measured. The fact that the asymptotic adaptation model fit significantly better than a sloped-line model suggests that adaptation occurred fast enough to be seen within the number of trials given.

The non-amputee subject's filtered perturbation data does not suggest any downward trend. It is quite possible that the subject was able to adapt to the change in the task so quickly that for the vast majority of the 80 perturbation trials, no significant adaptation was taking place.

It will be necessary to have several other amputee subjects participate in this experiment in order to make any general statements about the extent to which adaptation can be seen in amputees during perturbed crank turning.

7.2 Future Work

7.2.1 Showing Adaptation in Additional Subjects

For amputee subjects that follow, it will probably be helpful to shorten the crank handle, so that they will be able to turn the crank through a greater portion of its range. This will also help them to be able to turn the crank in their saggital plane. This will allow tasks to occupy a larger range of angles and will also allow the collection of meaningful force data.

Changing the nature of the perturbation may also make the task more appropriate. In future subjects, the perturbation will be modified to require a lower average force production from the amputee. The amount of force variation with position will be maintained or increased. The will decrease the amputee subjects' workload and reduce the possibility of amputee subjects experiencing discomfort during the experiment, while keeping the task challenging.

The data obtained from future amputee subjects may provide useful insights about amputees' motor- and prosthesis-control strategies, as well as serve as a possible basis for indirectly measuring adaptation.

7.2.2 Comparing Levels of Power-Assist

Once amputee adaptation in crank turning is established and better understood, it will be possible to examine the effect that the level of power-assist has on prosthesis performance. Amputee subjects can respond to perturbations while using both a high level of power assist and a low level of power assist. Adaptation patterns can be compared, showing to what extent the level of power-assist present affects prosthesis function.

Appendix A

Emulator Squeal

A.1 Summary

When a linear amplifier is attached directly to the emulator, the amplifier/emulator system oscillates at 4.4 kHz with a peak-to-peak voltage of 400 V and creates an audible squeal. The emulator squeal is due to amplifier saturation. The addition of a 45 μF bypass capacitor eliminates the emulator squeal and does not degrade emulator performance at frequencies lower than 100Hz.

A.2 Background

A.2.1 Hardware description

The emulator is powered by a Kepco Bipolar Operational Amplifier (a linear amplifier) operating in current control mode. When in current control mode, the amplifier has access to a voltage supply of $\pm$200 V that it can use to drive the commanded current through a given circuit. Whenever a voltage of magnitude greater than 200 V is required to drive the requested current, then the amplifier is in a state of saturation.

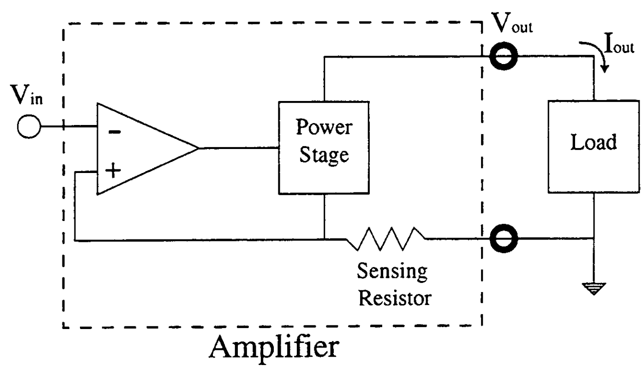


Figure A-1: Simplified schematic of the amplifier current control feedback loop.

The current control circuit in the amplifier is regulated with feedback. An input voltage of ± 10 V commands the amplifier over its entire range of ± 4 A. Current is controlled through a simple feedback loop (see figure A-1) A sensing resistor is placed in the current path, and the voltage drop across it is fed back into the input.

A.2.2 Squeal description

When the amplifier is attached directly to the emulator, the amplifier/emulator system oscillates at 4.4 kHz, regardless of the input voltage (even when the input is grounded). The oscillations have a peak-to-peak voltage of 400 V and create an audible squeal (see figure A-2).

When viewed with an oscilloscope, the squeal appears similar to a square wave except that its peaks slope towards 0 V, rather than remaining flat.

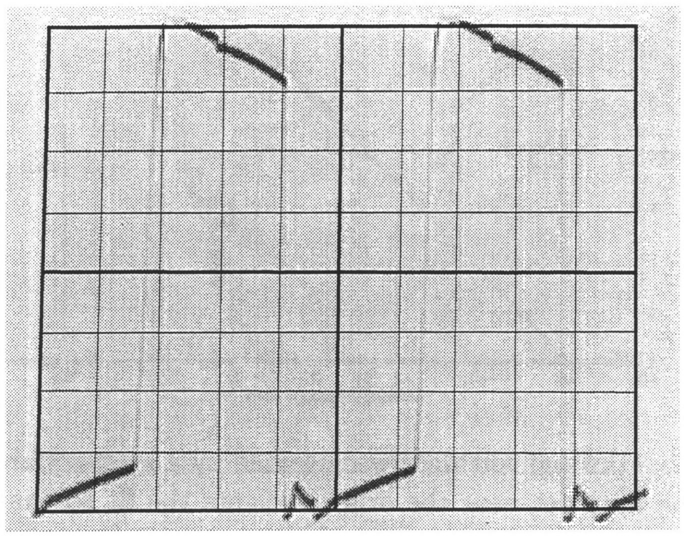


Figure A-2: Emulator squeal. Oscilloscope set to 50 V/div and 50 ms/div. This is an amplifier output voltage measurement taken with an oscilloscope during squeal. The input voltage is 0 V.

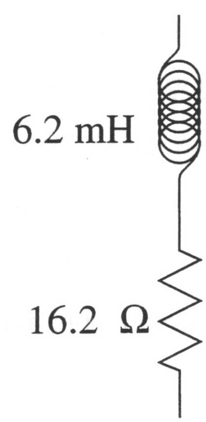


Figure A-3: Electrical model of the emulator's DC motor.

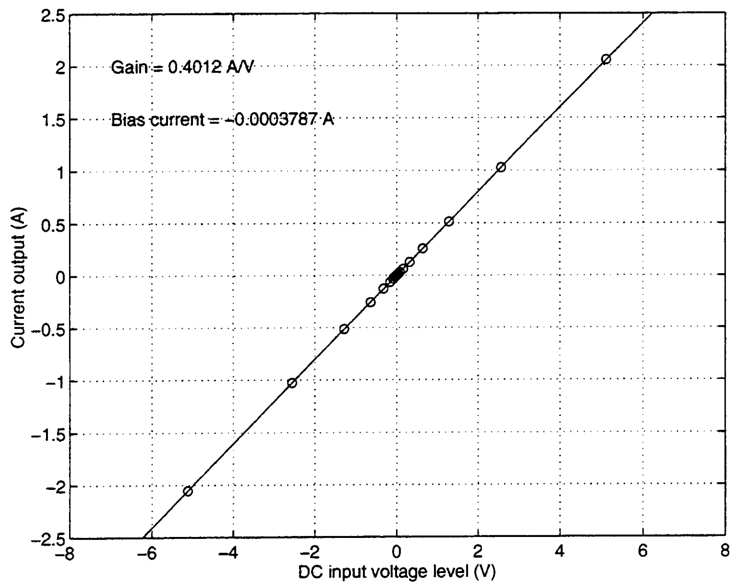


Figure A-4: The line through the data was created by a least-squares algorithm. The high density of the data set near the origin weights the fit so that it is best at the origin.

A.3 System Modeling

A.3.1 DC Motor Model

In order to better understand the squeal, I first had to model the DC motor and the amplifier. The electrical impedance of the emulator's DC motor can be modeled as an inductance in series with a resistance (see figure A-3). The measured resistance of the emulator with its power cable is 16.2 Ω. When modeled, an inductance of 6.2 mH best predicts the emulator's frequency response. The manufacturing specifications list an inductance of only 3.8 mH. This discrepancy has not been investigated, rather, the value of 6.2 mH has been chosen for use in the model and, as will be shown later, produces a model that predicts the system behavior well.

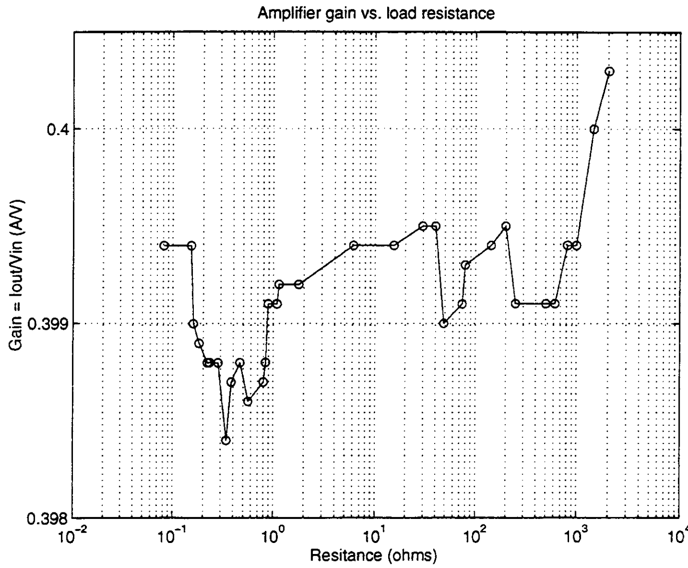


Figure A-5: Change in amplifier gain with variations in load resistance.

A.3.2 Amplifier DC Gain

Nominally, the amplifier produces 0.4 A/V over its operating range. To verify this, I varied the input voltage from -5.12 V to +5.12 V while measuring the current with a precision ammeter (see figure A-4). The line that best fits the data has a slope of 0.4012 A/V with a y-intercept of 0.4 mA. This indicates that the gain of 0.4 A/V is accurate to about 1%.

In order to check the dependence of the gain on load resistance, I held the input voltage constant while substituting load resistors ranging from 0.07 Ω to 2.0 kΩ (see figure A-5). All measured values were within 1% of the mean. In fact, for resistances less than 500 Ω , measured values were within 0.5% of the mean. This shows that the gain is reasonably constant for a range of resistive loads.

A.3.3 Amplifier Dynamics

I determined the frequency response of the amplifier by placing a non-inductive resistor (metal film or carbon type) as the load, setting the amplitude of the input voltage,

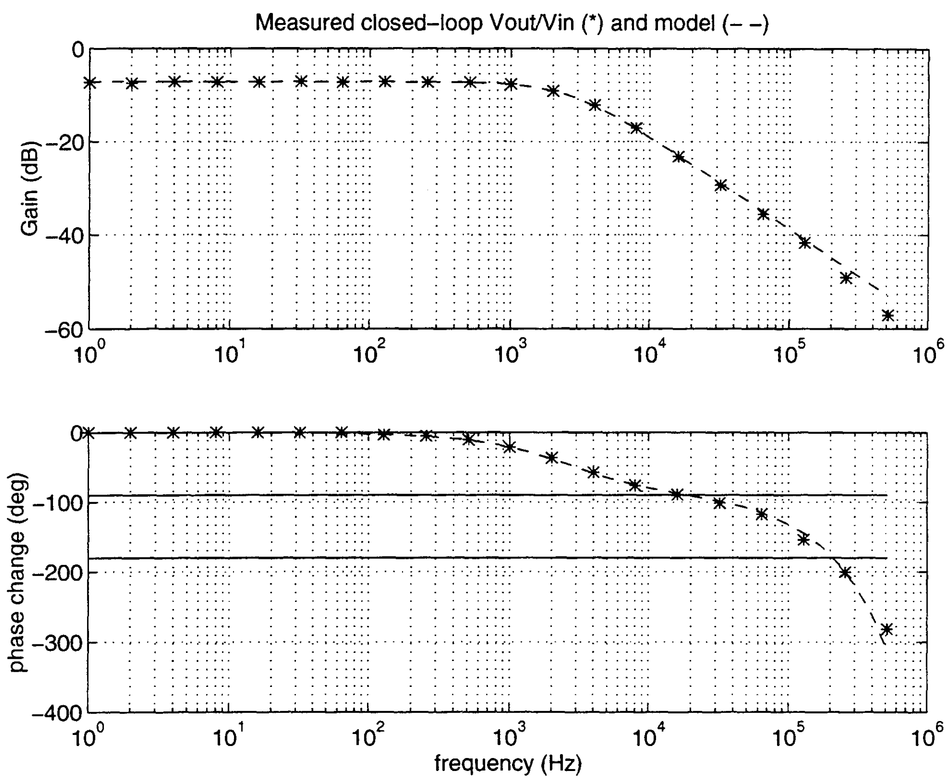


Figure A-6: Frequency response (Vout/Vin) of the amplifier with a resistive load. Measured data points are marked '*'. The dotted line shows behavior predicted by the model. Plots similar to this were obtained with various resistive loads, varying from 0.07 Ω to 55 Ω. In every case, the break frequency occurred at 2.6-2.7 kHz.

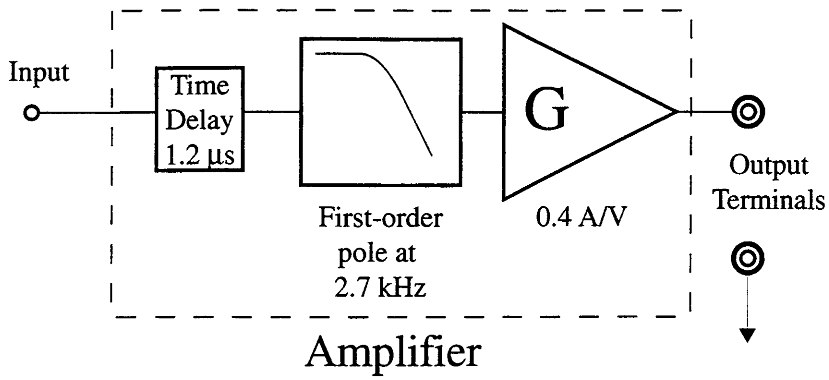


Figure A-7: Block diagram model of the amplifier.

and varying its frequency while measuring the output voltage and phase difference. This allowed the construction of Bode-type plots. Resistance values ranged from 0.07 Ω to 55 Ω. In each case the frequency response was nearly identical. The figure shown is typical (see figure A-6). The frequency response can be modeled by a first-order pole at 2.7 kHz and a 1.2 μs delay.

A.3.4 Amplifier Model

The DC gain and frequency response information allows the construction of a model that describes the amplifier's behavior (see figure A-7). The output impedance of the of the amplifier is sufficiently low that the amplifier dynamics do not change significantly over the range of loads considered. For this reason the first-order pole in the amplifier can be modeled as a block in the block diagram.

A.4 Squeal Investigation

A.4.1 Squeal is an electrical phenomenon

I considered the possibility that the squeal is a resonant vibration of a mechanical component, such as the brushes in the DC motor. However, when a resistor and

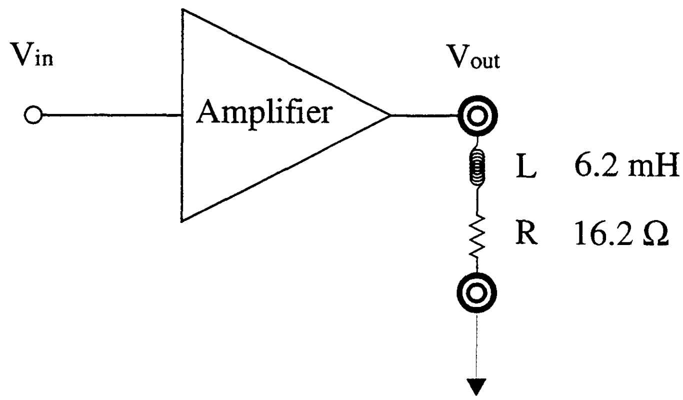


Figure A-8: Schematic model of the amplifier/emulator system.

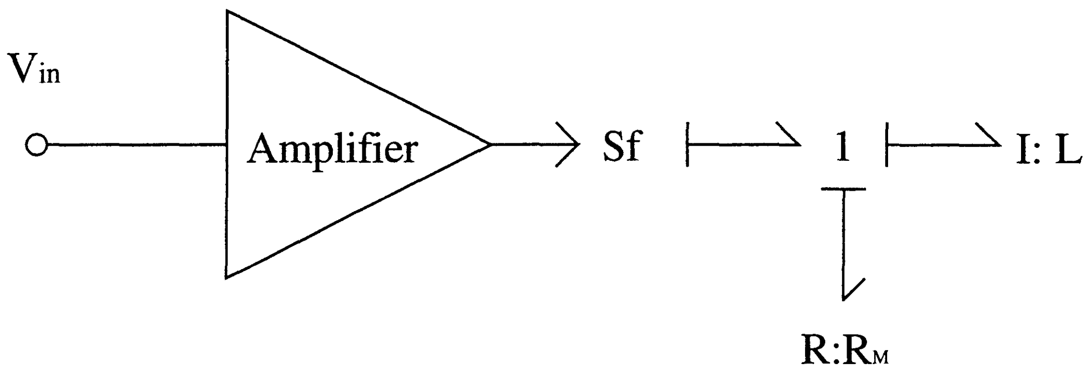


Figure A-9: Bond graph model of the amplifier/emulator system. Note that the inductance of the motor is subject to derivative causality.

inductor are attached to the amplifier they also produce a squeal, even though they are not attached to any mechanical components. In addition, the squeal rises in frequency when a smaller inductor is used. Because it is possible to recreate the squeal in the absence of any mechanical hardware, and because its frequency varies with the inductance of the load, it is clear that the squeal is an electrical phenomenon rather than a mechanical resonance.

A simulation of the amplifier/emulator system shows an interesting feature. The two models are connected in this way (see figure A-8). When the system is stated in bond graph form, equations that take into account physical interaction between components can be generated (see figure A-9). The current source in the bond graph forces the inductor to have derivative causality. This means that, no matter how high

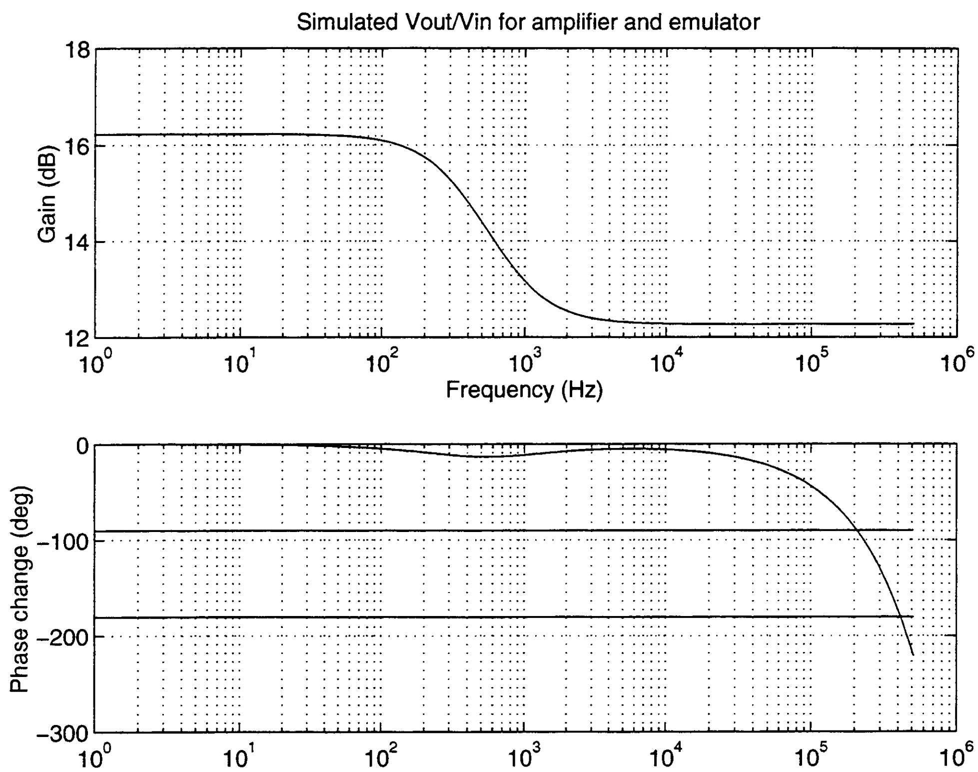


Figure A-10: Simulated frequency response (Vout/Vin) of the amplifier/emulator system.

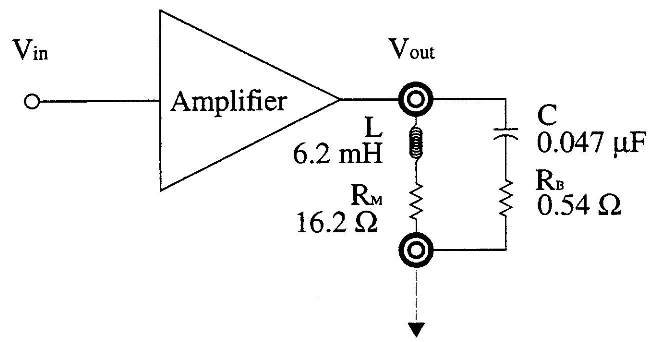


Figure A-11: Model of the amplifier/emulator system with a bypass capacitor. L is the motor inductance, R_M is the motor resistance, C is the bypass capacitance, and R_B is the equivalent series resistance of the bypass capacitor.

the load's impedance gets, the current source will force a fixed amount of current through it. So when the impedance gets large (as an inductor's impedance will at high frequencies), the output voltage simply gets large in proportion.

When simulated, the system produces the Bode plot in figure A-10. This model does not predict saturation. It does have a gain of 12 dB when the phase drops below 180 degrees, which would predict instability. However, there have been no tests of the complete system to see whether the physical model is valid in that frequency range ($> 10^5$ rad/sec).

A.4.2 Squeal is due to amplifier saturation

In order to alleviate the squeal, a Kepco engineer recommended placing a bypass capacitor across the power output leads. When a 0.047 μF capacitor is added across the amplifier output leads (see figure A-11), the emulator no longer squeals at most frequencies.

Figure A-12 shows the response of the amplifier driving the emulator and bypass capacitor load at a commanded current amplitude of ± 0.4 A. Note the relation of the amplifier saturation level to the squeal region.

The same system being driven at ± 0.12 A produces the frequency response shown

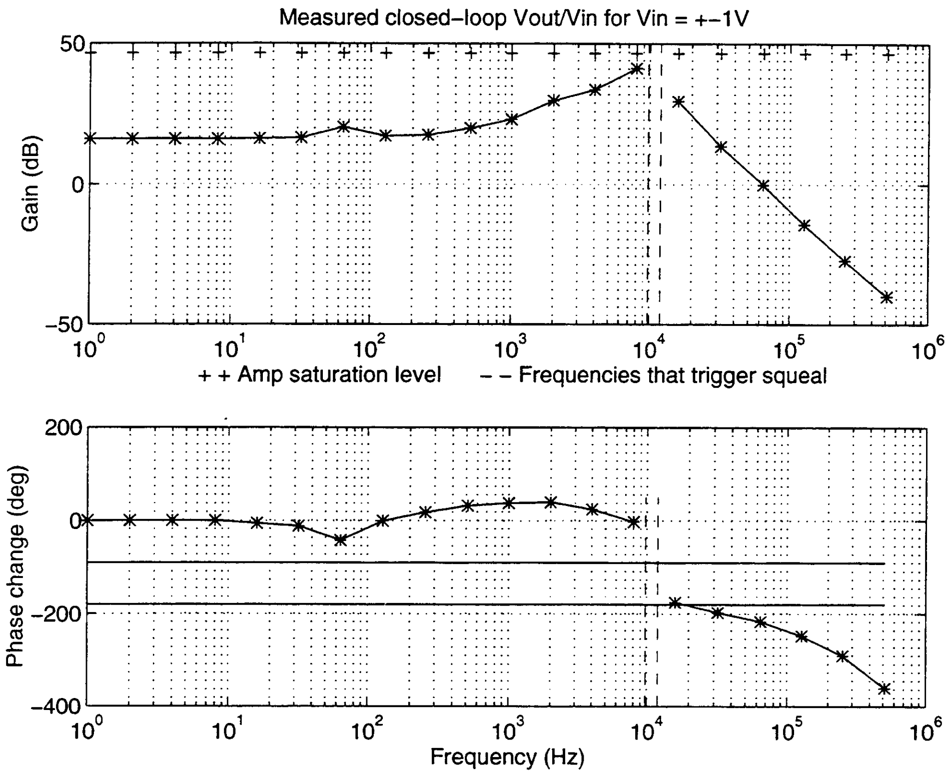


Figure A-12: Measured frequency response (Vout/Vin) of the amplifier/emulator system with a 0.047 μF bypass capacitor. Input is a sine wave of amplitude 2 V. Measured data points are marked '*'. Note that the squeal region begins and ends roughly where Vout/Vin crosses the saturation level.

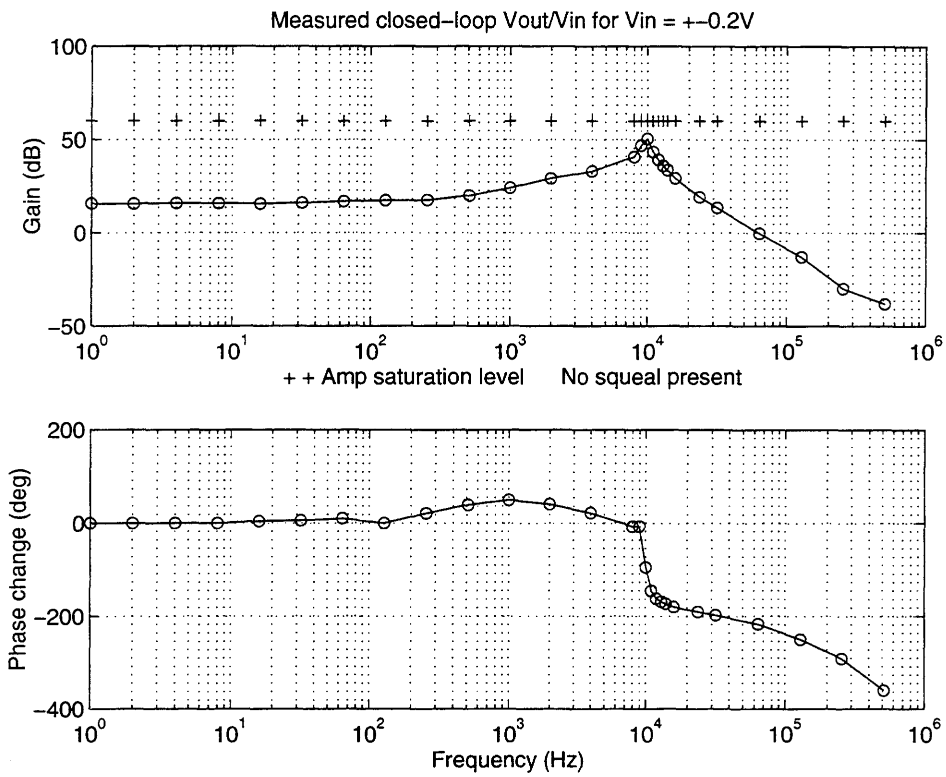


Figure A-13: Measured frequency response (Vout/Vin) of the amplifier/emulator system with a 0.047 μF bypass capacitor. Input is a sine wave of amplitude 0.4 V. Measured data points are marked 'o'.

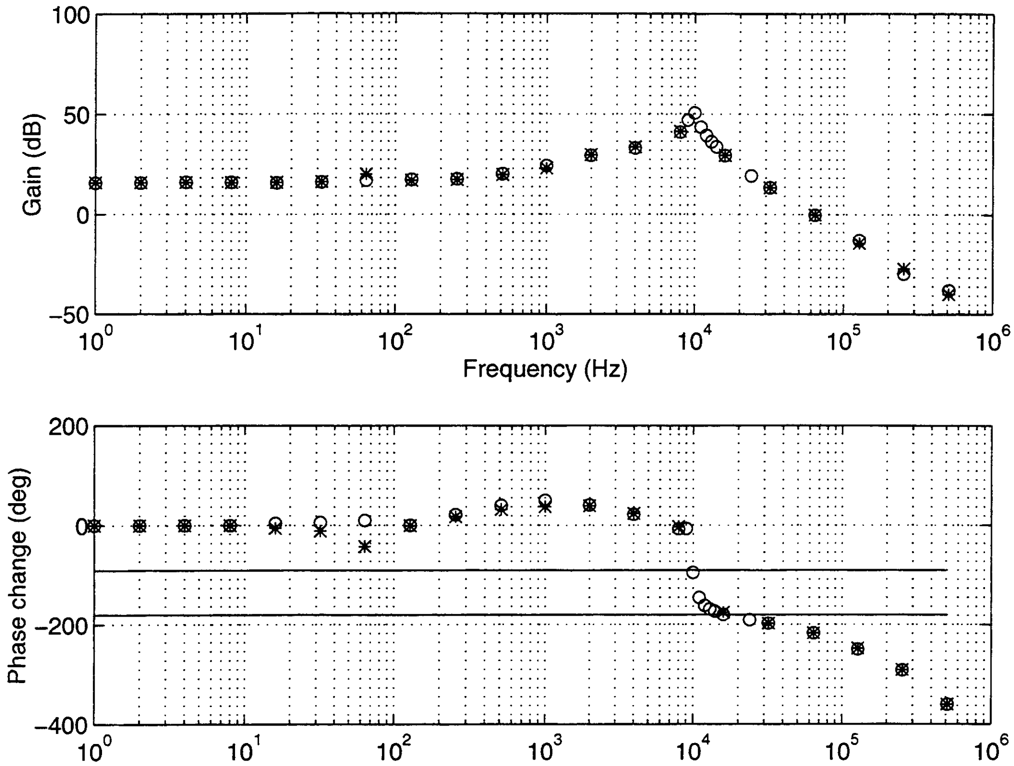


Figure A-14: Frequency response data (Vout/Vin) for both high and low input voltage levels superimposed

in figure A-13. Note the correspondence between the change in amplifier saturation level and the disappearance of the squeal region.

The two sets of data overlap nearly exactly with the exception of the squeal region. In the vicinity of the squeal region the system is behaving nearly identically at both input levels (see figure A-14). This gives strong evidence that the system is behaving linearly and that the squeal in this case is not caused by a linear instability. From these data, it seems clear that the squeal is a result of the amplifier's saturation behavior.

A.4.3 Discussion of amplifier saturation

It is important to note, however, that the saturation behavior varies with a change in load. When large resistive loads are used, the saturated amplifier simply drives

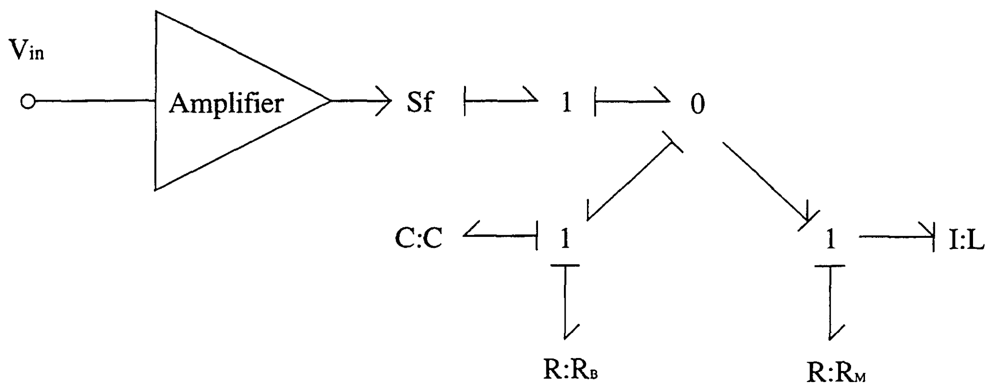


Figure A-15: Bond graph of the amplifier/emulator system with a bypass capacitor.

the output voltage to the rail. Causing the amplifier to saturate does not in and of itself produce squeal. Squeal is apparently a product of the interaction between the saturated amplifier and an inductive load.

A.5 Analysis of the Addition of the Bypass Capacitor

A.5.1 System Model with the Bypass Capacitor

When measured with an impedance analyzer, the 0.047 μF capacitor displays an actual capacitance of 0.0403 μF and a series resistance of 0.321 Ω. It can be added to the amplifier/emulator bond graph model as shown if figure A-15. The addition of the bypass capacitor allows the inductor to assume integral causality.

When a plot of the model's prediction of system behavior is superimposed on the measured data, it shows a reasonably good fit (see figure A-16). The only significant discrepancy is a small resonance at about 65 Hz. At this frequency the emulator shook noticeably, presumably due to a mechanical resonance. However, no effort has yet been made to characterize or model this behavior.

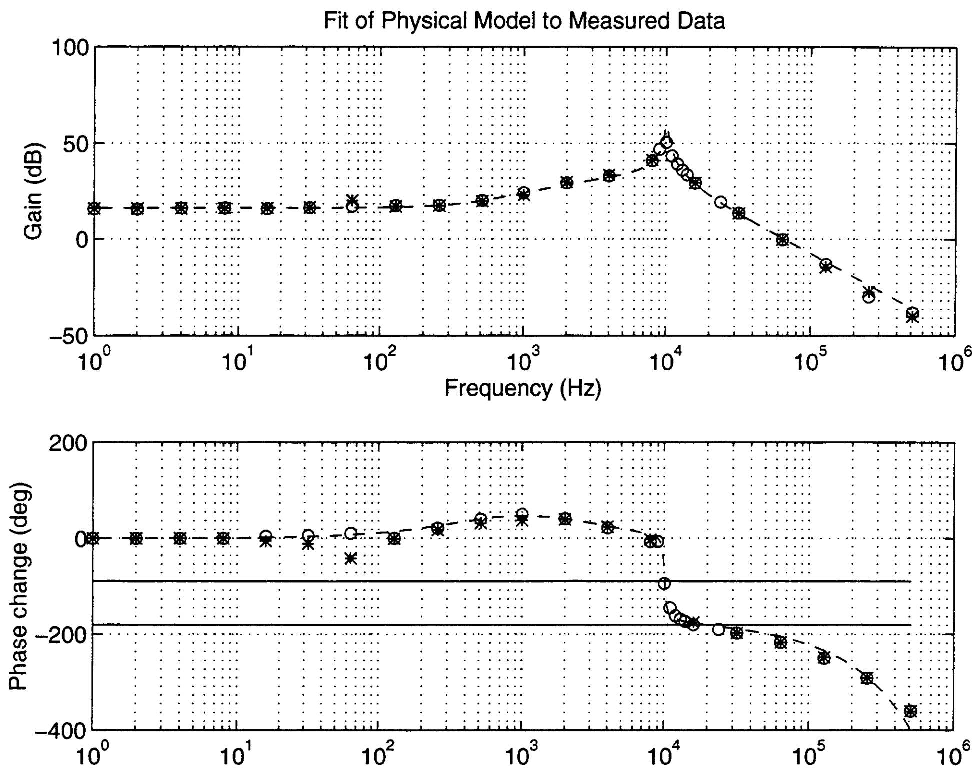


Figure A-16: Frequency response of the amplifier/emulator system with the bypass capacitor in place. Measured data and model prediction of Vout/Vin superimposed.

A.5.2 Choosing an Ideal Capacitance Value

Ideally, the amplifier will have the following characteristics:

1. Total circuit impedance never exceeds 50 Ω. This way, even when driving the amplifier's maximum current capacity of 4 A, the output voltage never exceeds 200 V.

2. Motor current does not vary by more than 3 dB from 0-100 Hz. This is a conservative estimate of the maximum meaningful control signal bandwidth expected from the amputee.

3. Motor current does not lag voltage input by more than 45 degrees below 100 Hz.

In simulations, these criteria are easily met by increasing the resistance of the bypass circuit. 450 Ω turns out to be a good value for minimizing overall load impedance. In practice, however, this puts tremendous power dissipation requirements on the resistor, and unfortunately all readily available high-power resistors are wire-wound resistors, which add a significant and undesirable inductive load to the bypass circuit. Increasing the bypass circuit resistance by only a few ohms does not appreciably affect the overall circuit impedance. That limits the problem to optimization of the capacitor.

Increasing the capacitance value decreases the resonant frequency of the load. It also decreases the height of the resonant peak in the impedance plot (see figure A-17). The optimum capacitor, then, is the largest one that does not excessively influence motor current below 100 Hz. A 50 μF capacitor accomplishes this well. The total circuit impedance never rises much above the 16.2 Ω resistance of the motor, and therefore, the voltage would never exceed 65 V. This would effectively remove any possibility of recurring squeal. In addition, this would protect any components that may be sensitive to high voltages.

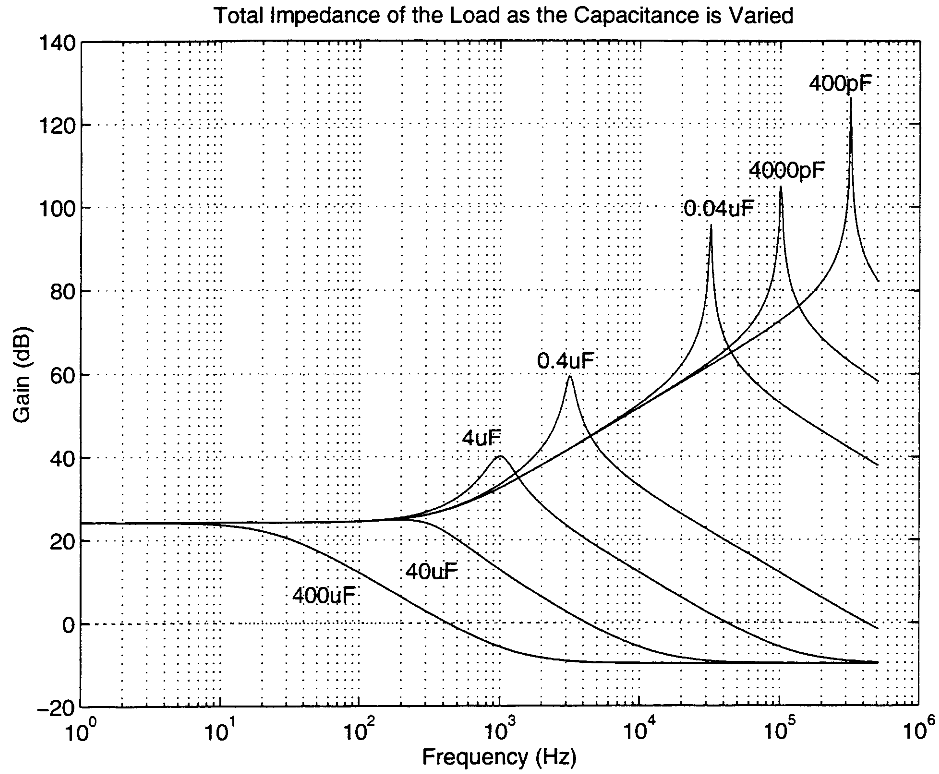


Figure A-17: Load impedance as function of bypass capacitance. 0 dB = 1 Ω.

The resulting bandwidth of the motor current is acceptable (see figure A-18). It drops by 3 dB only at frequencies higher than 200 Hz, and its phase exceeds -45 degrees near 150 Hz.

When tested in hardware, a 45 μF capacitor causes the circuit to behave much the same way the model predicts (see figure A-19). The only major discrepency is the small peak near 70 Hz that was noted in the discussion of figure A-16.

The closeness of the fit is confirmation that the model used is a reasonable description of the physical interactions that are taking place within the frequency range tested.

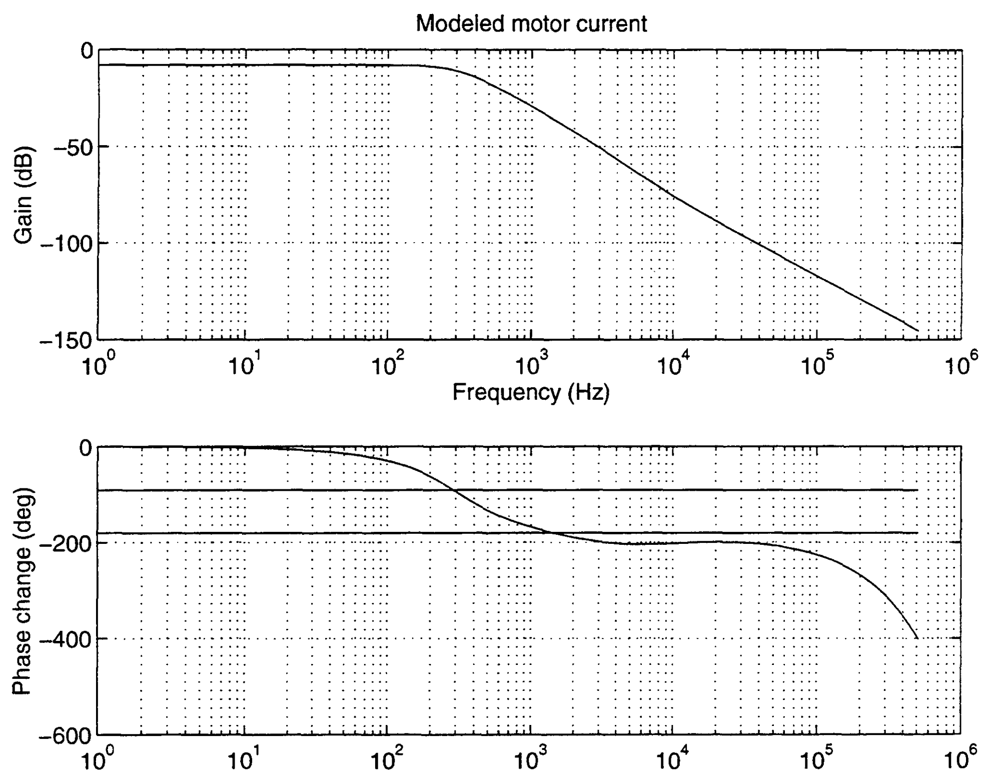


Figure A-18: Modeled frequency response of motor current with a 50 μF bypass capacitor.

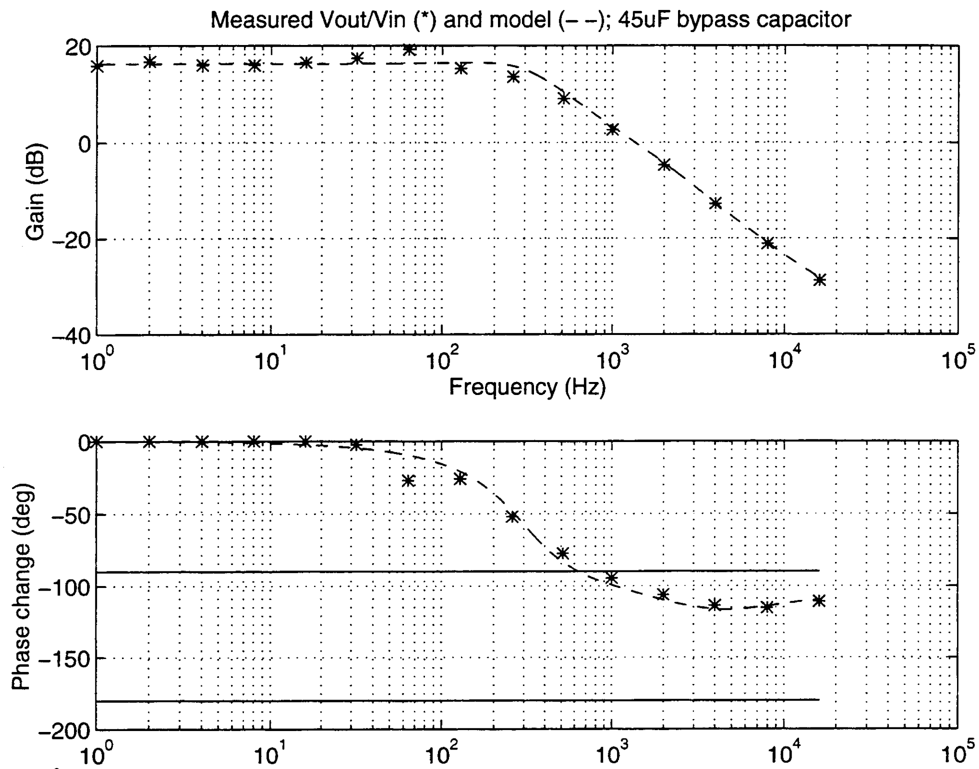


Figure A-19: Measured data and model prediction of frequency response (Vout/Vin) of the amplifier/emulator system with a 45 μF bypass capacitor.

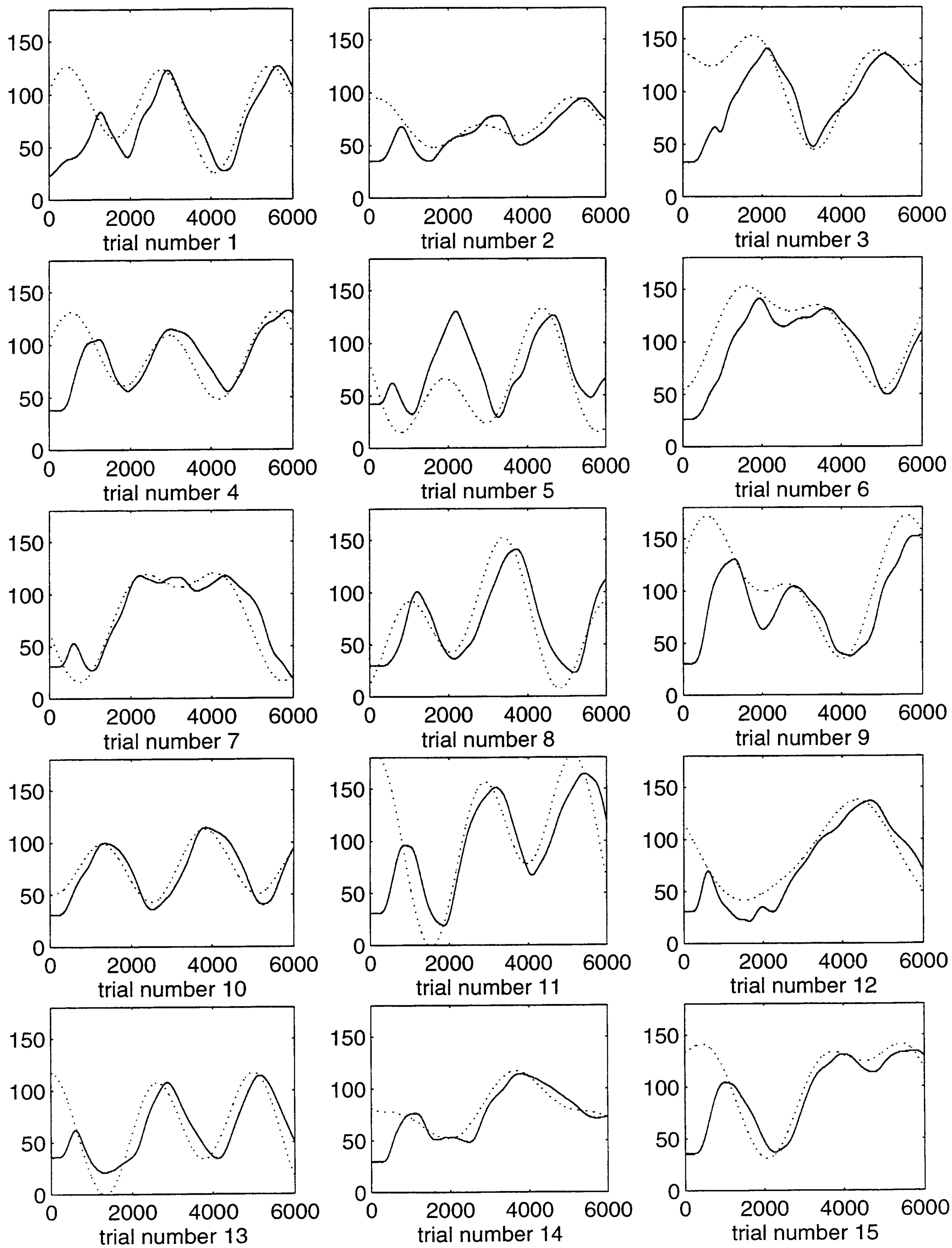
A.6 Conclusions

Emulator squeal is electrical in nature. The addition of a 45 μF bypass capacitor eliminates the squeal without significantly changing the magnitude of current to the motor for input frequencies below 100 Hz.

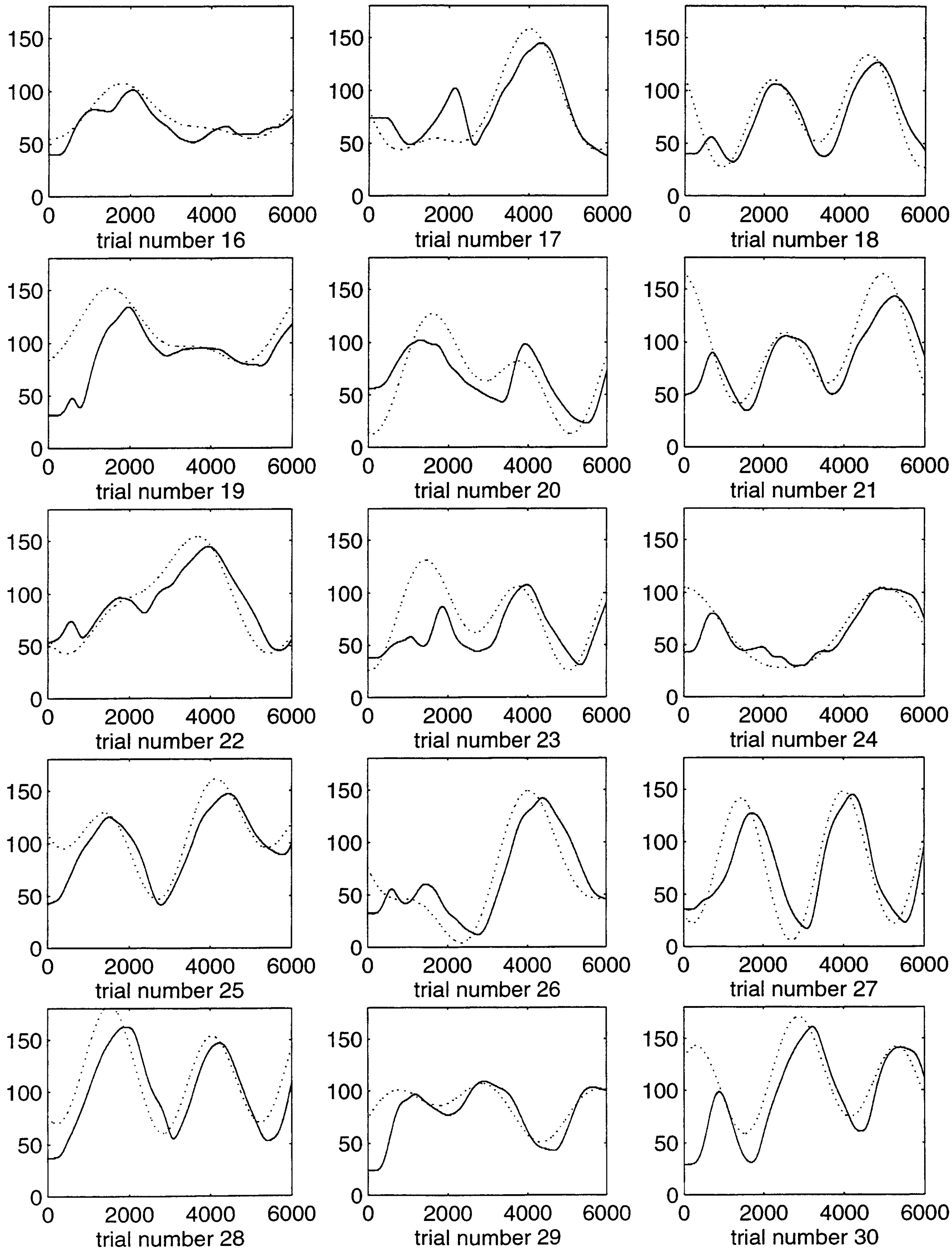
Appendix B

Data from the Non-amputee Subject

Data Plots: Pre-Perturbation Trials



Data Plots: Pre-Perturbation Trials



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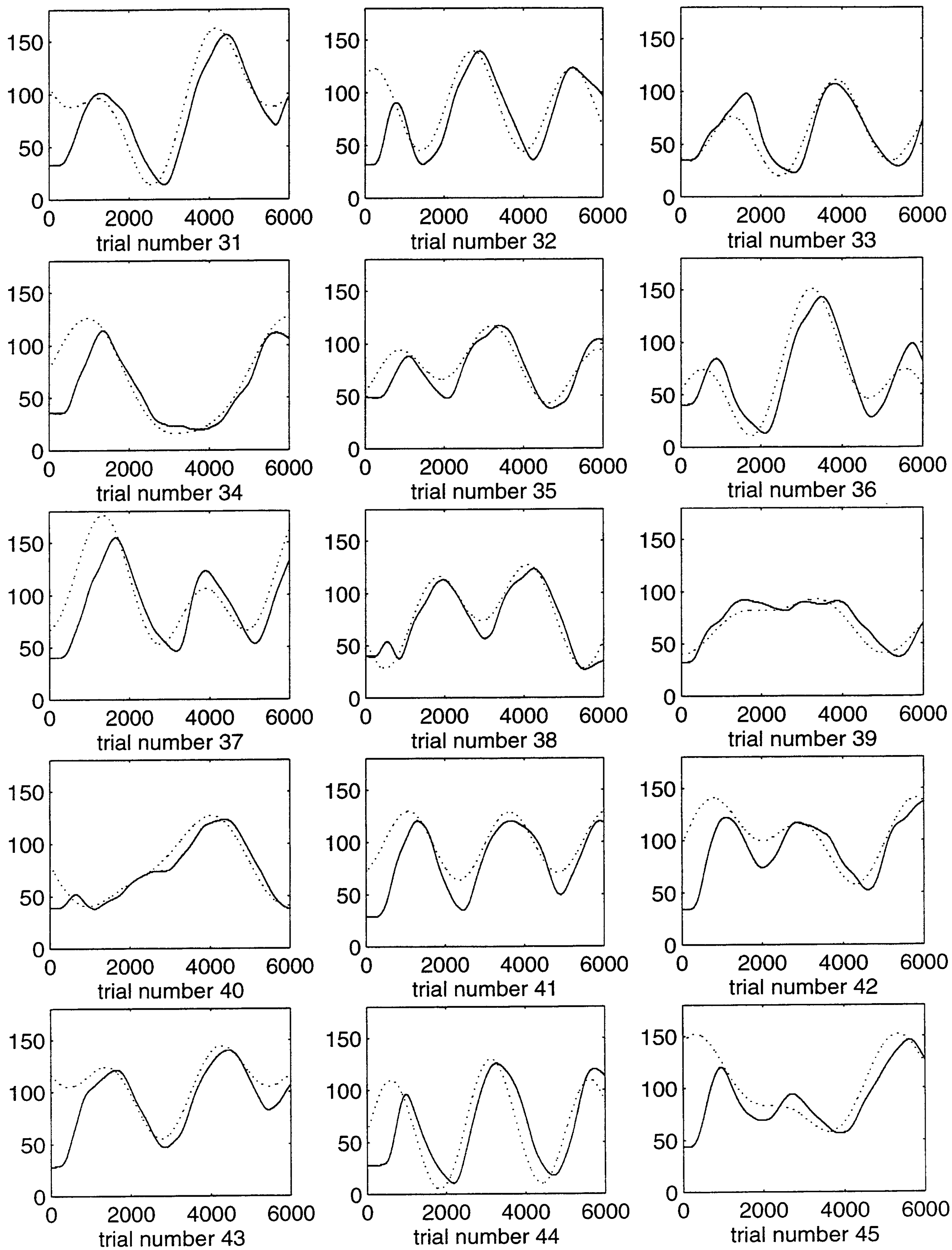

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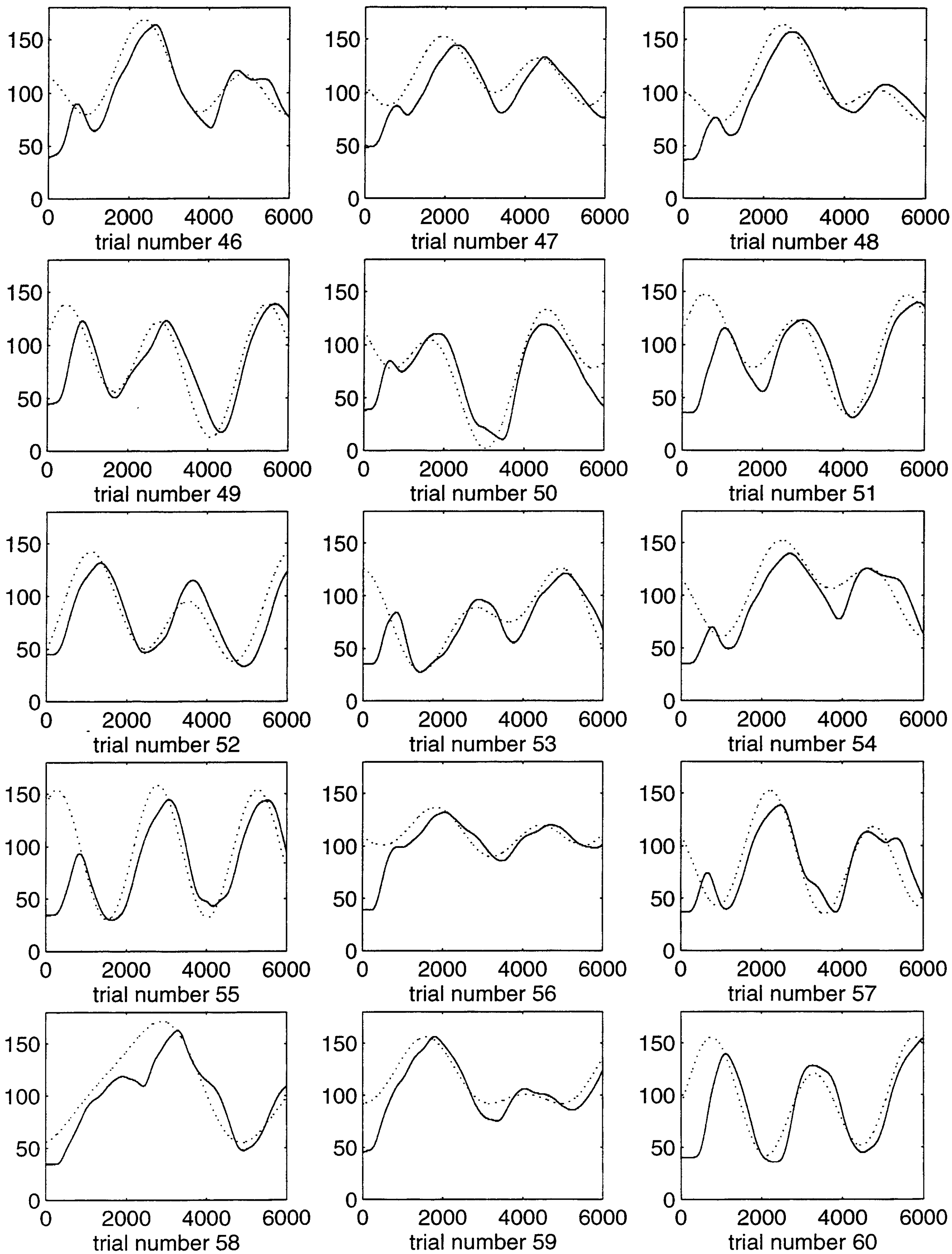

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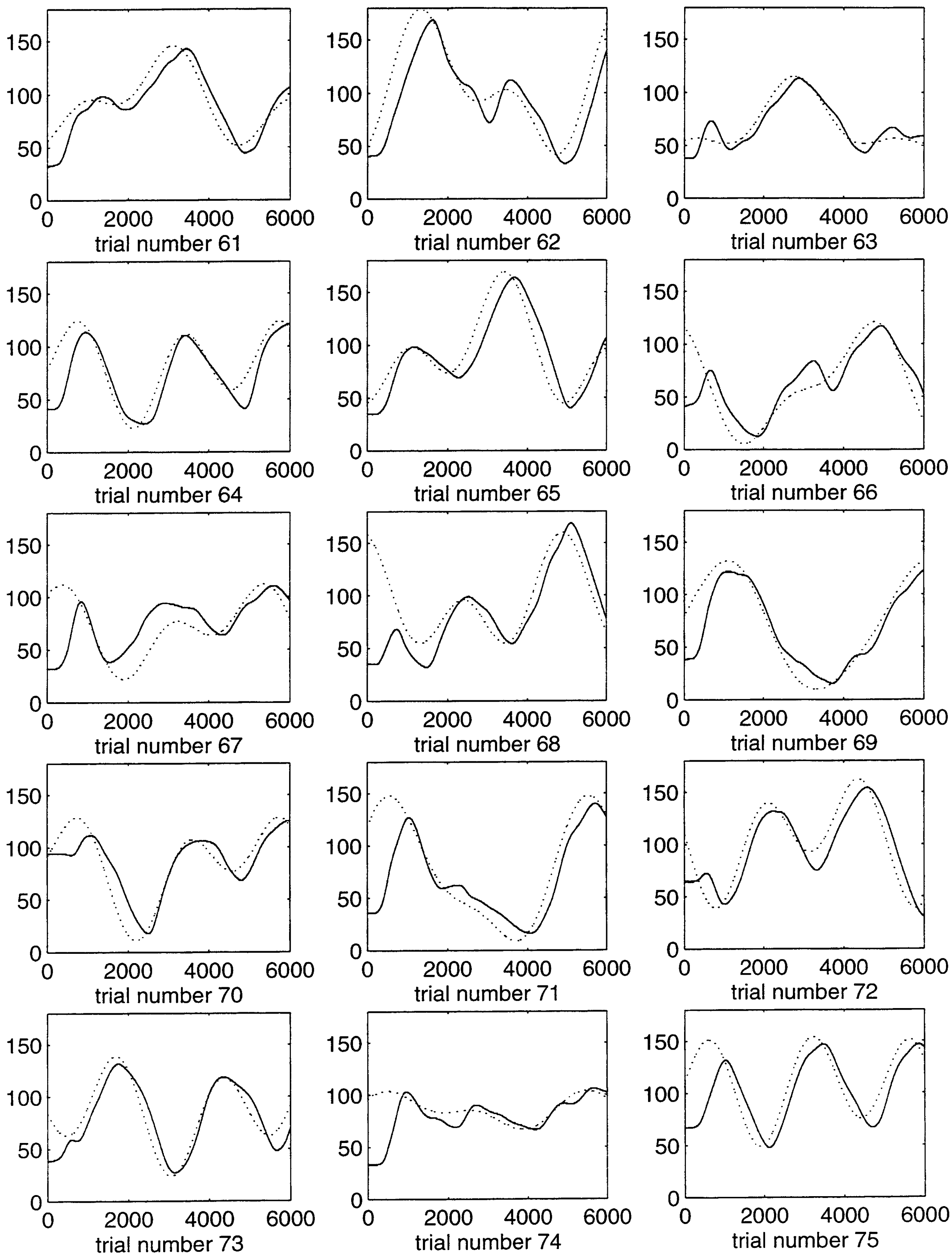

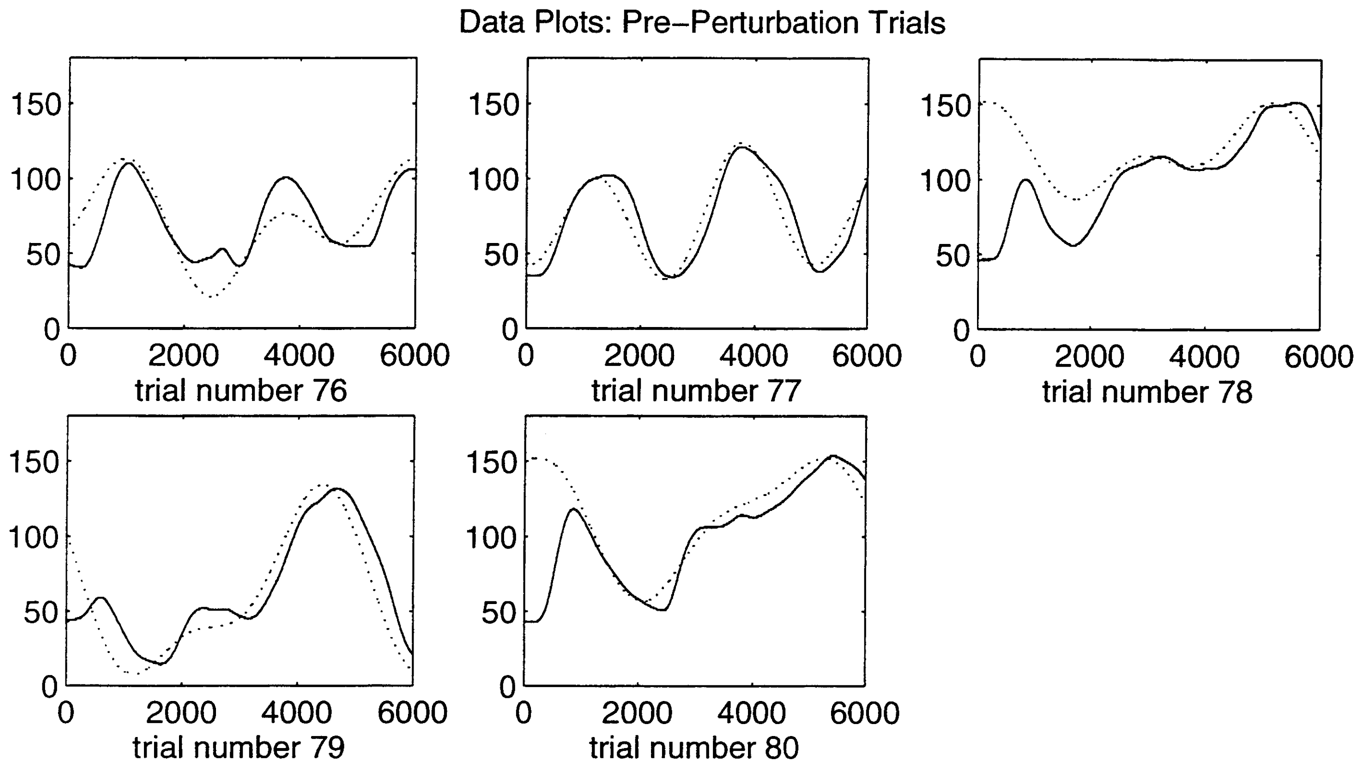
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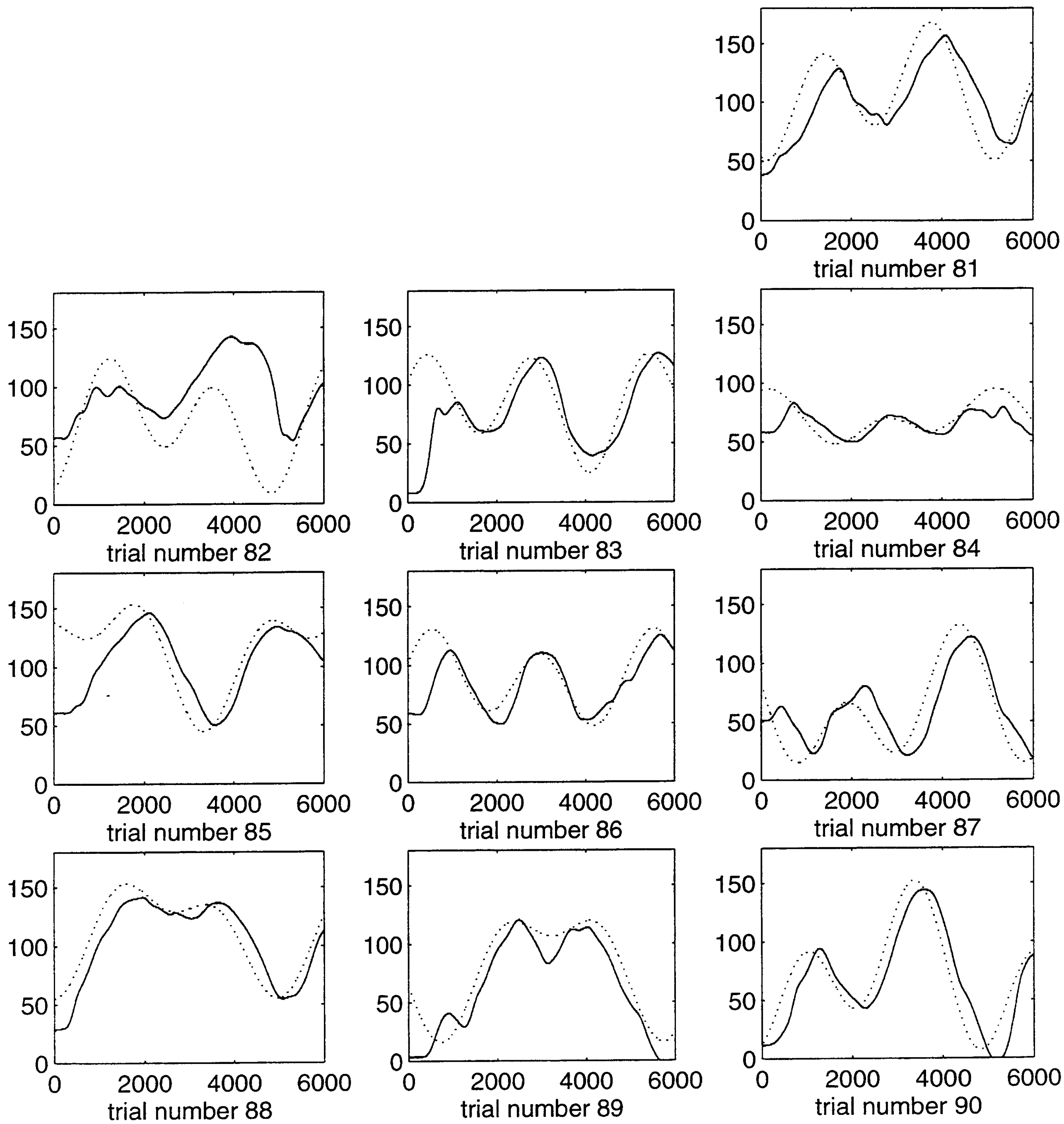
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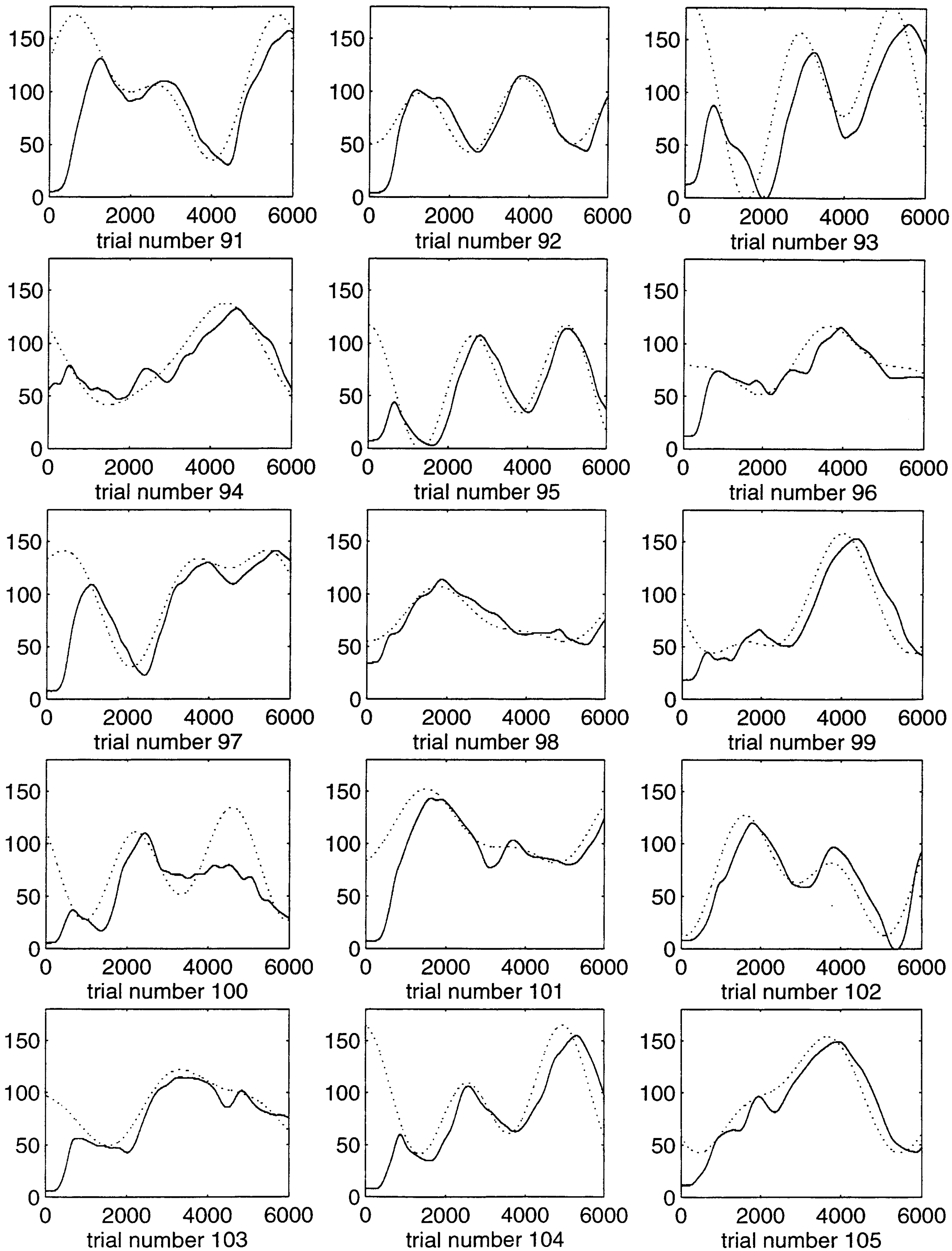
Data Plots: Perturbation Trials



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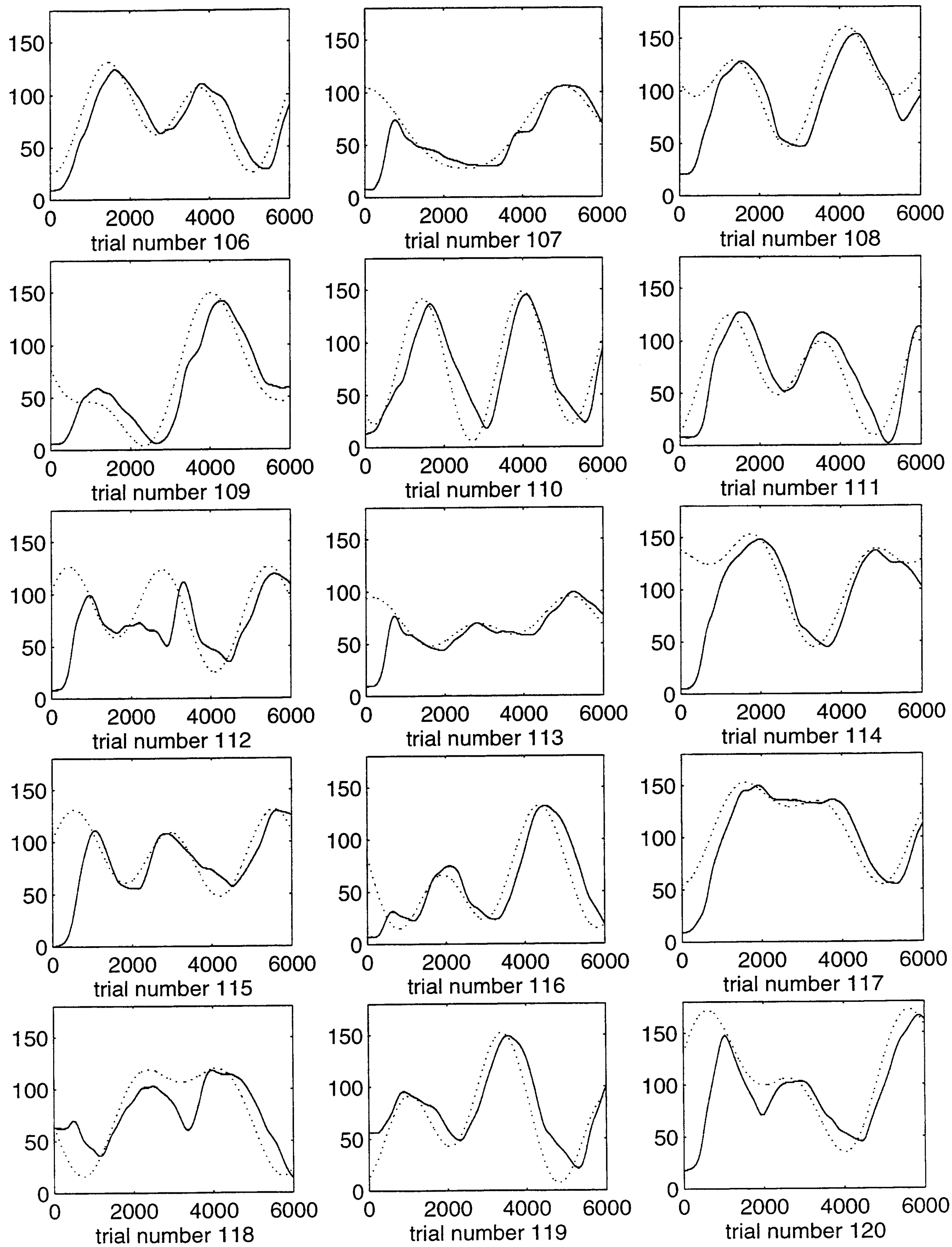

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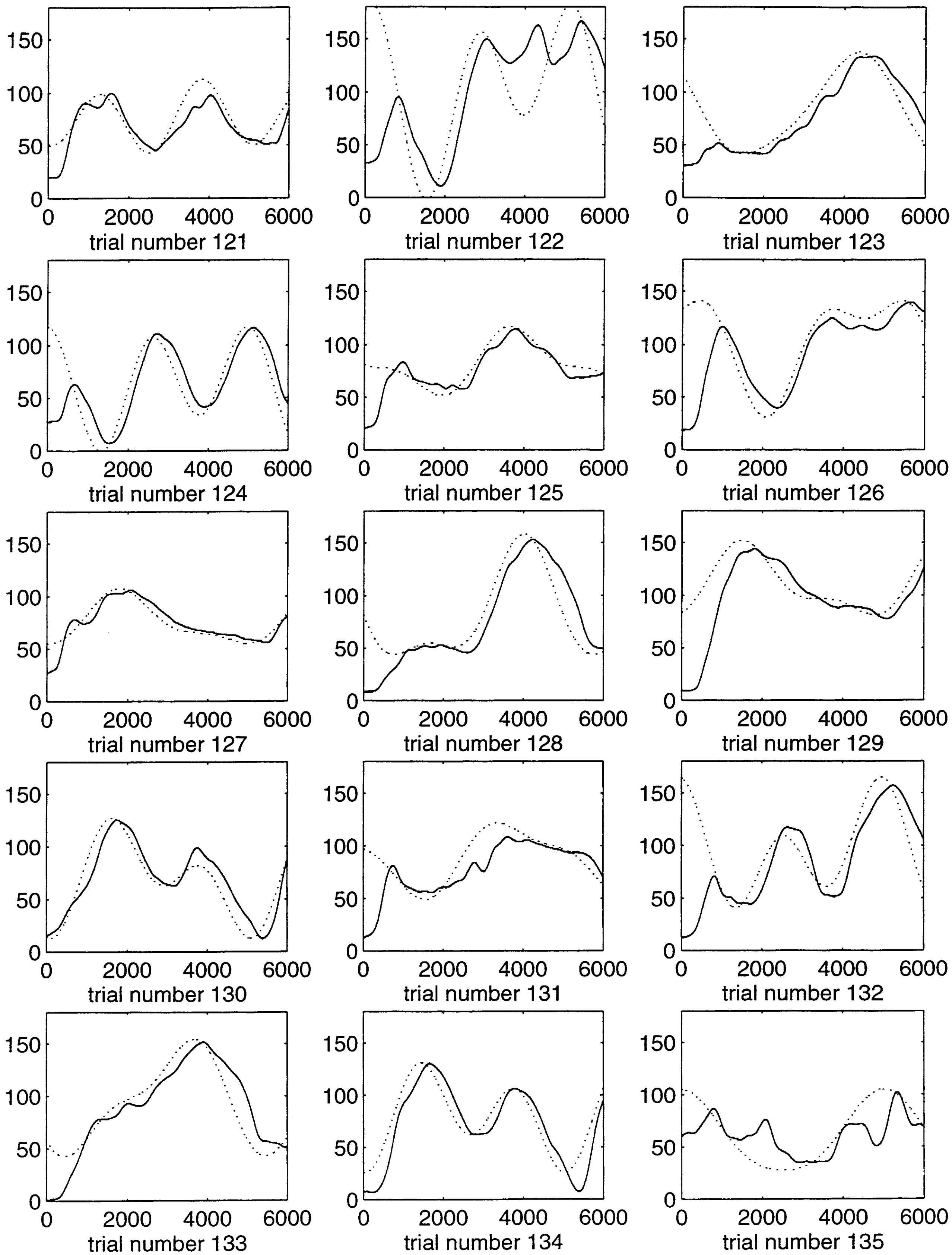

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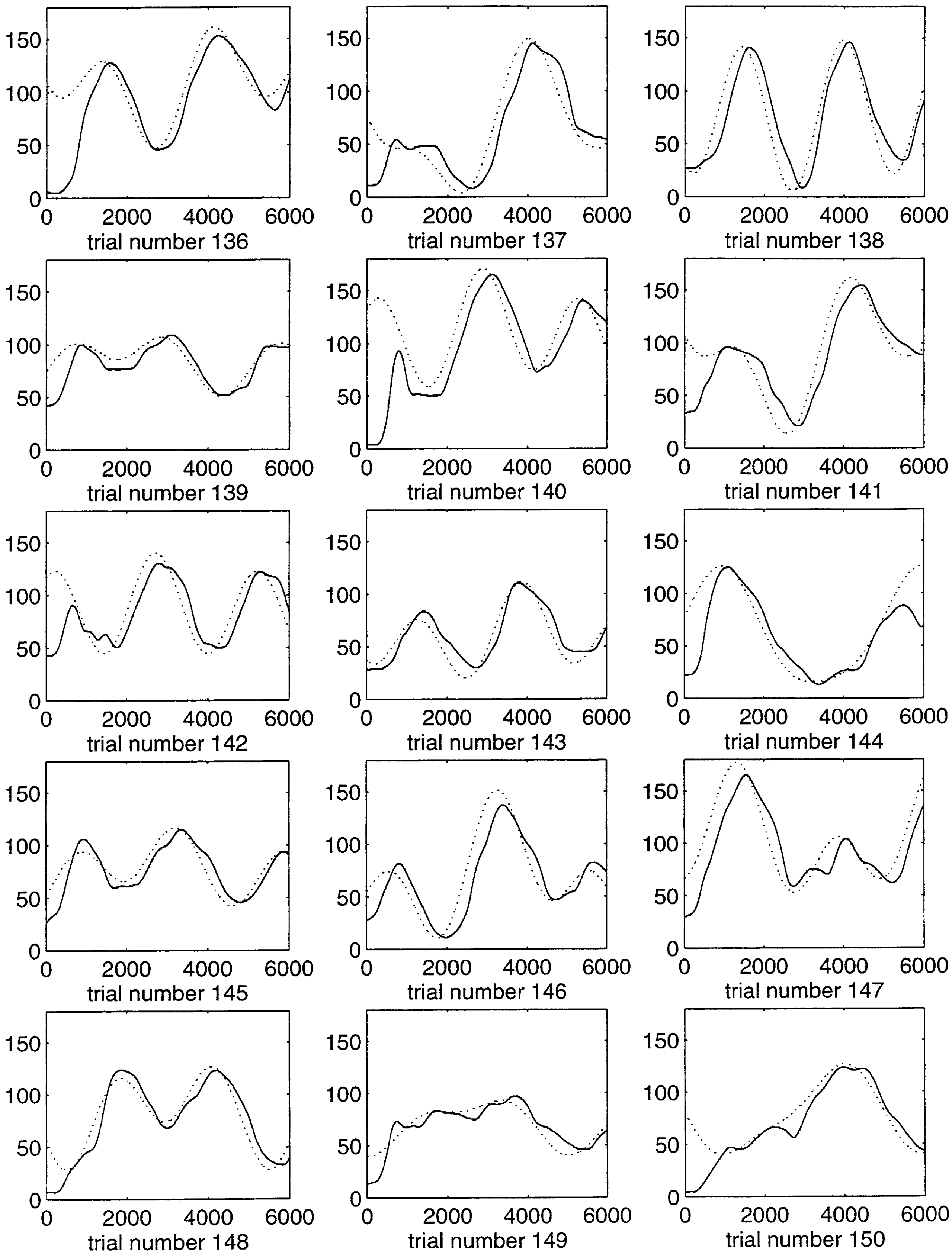

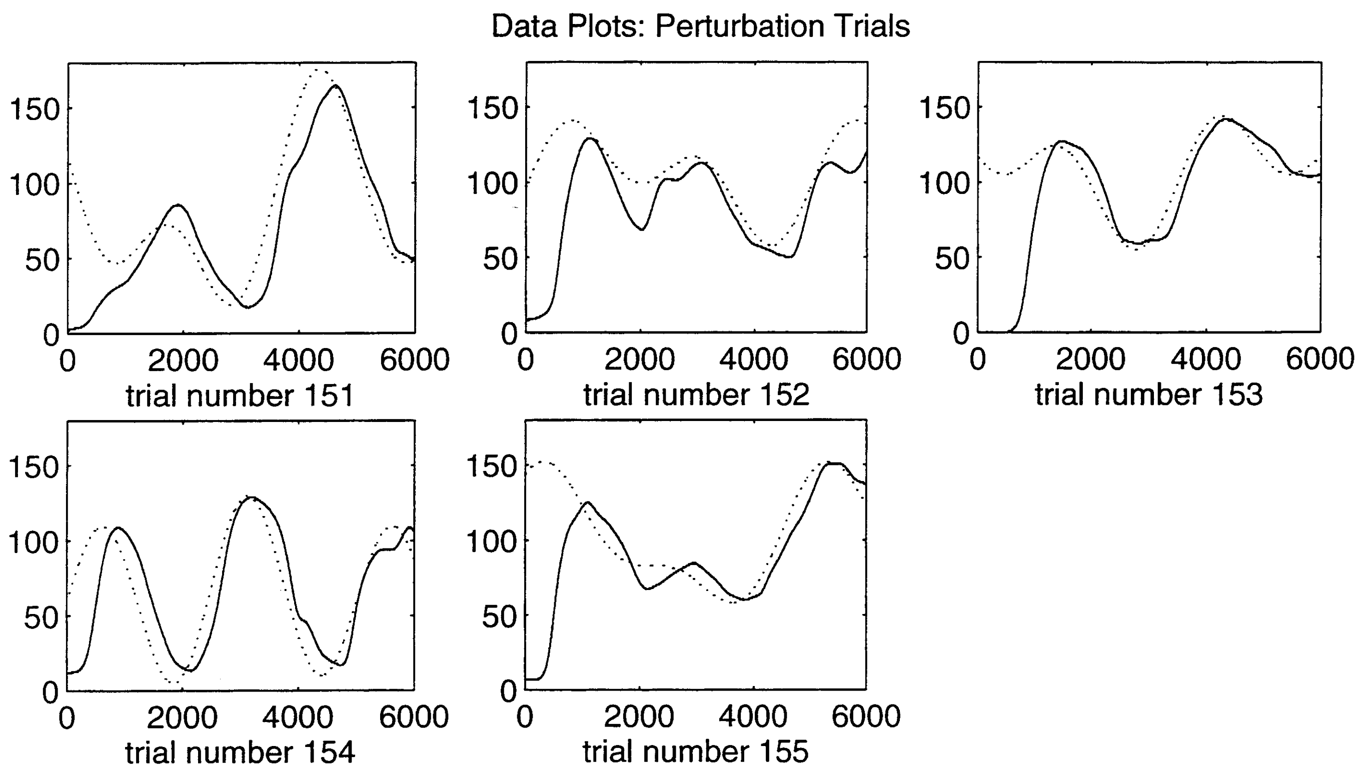
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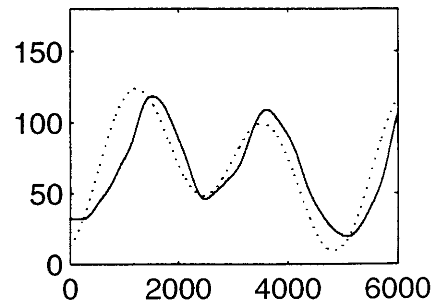

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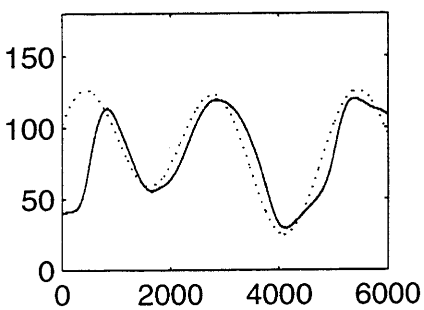

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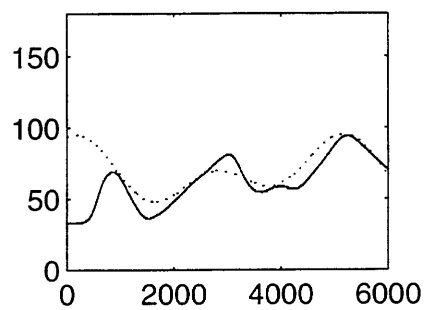

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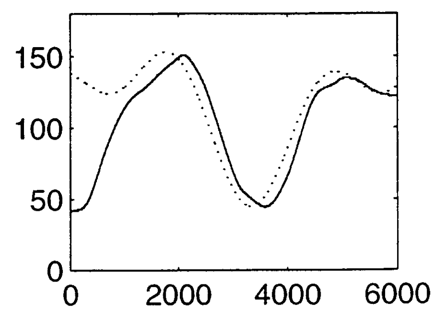

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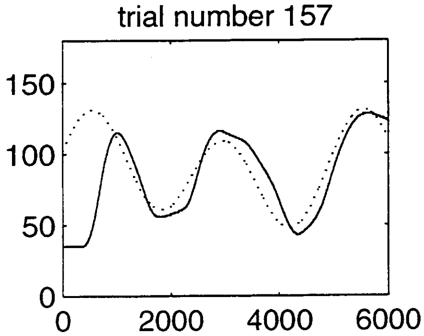

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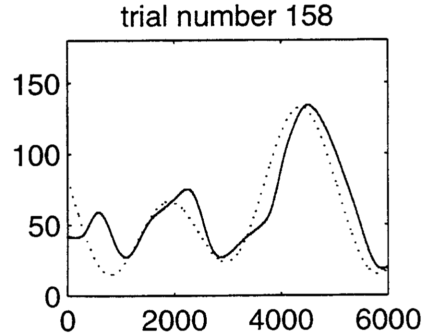

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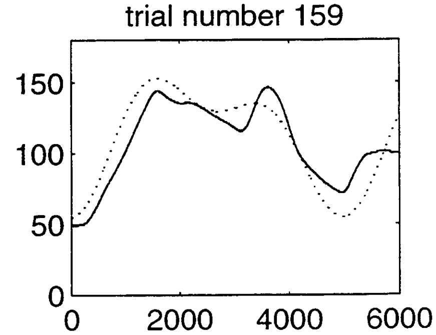

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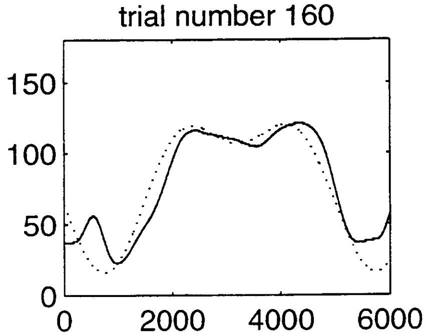

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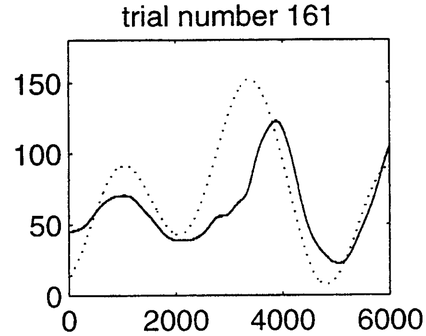

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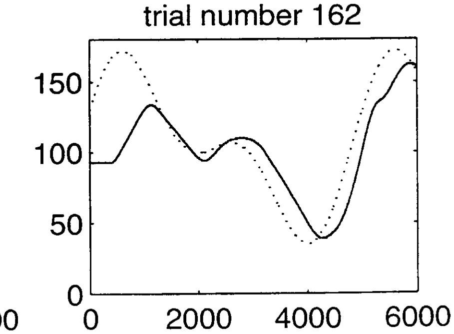

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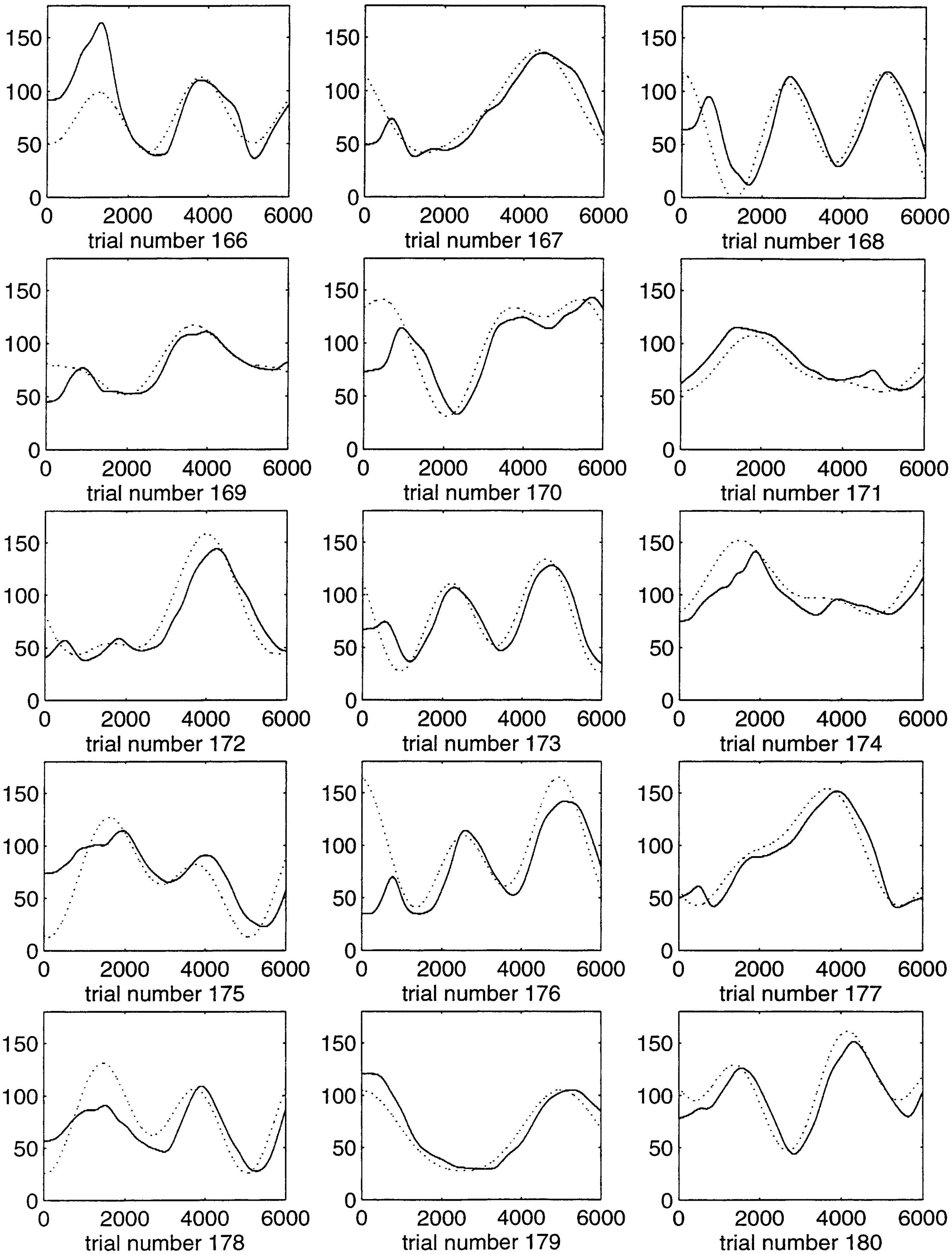
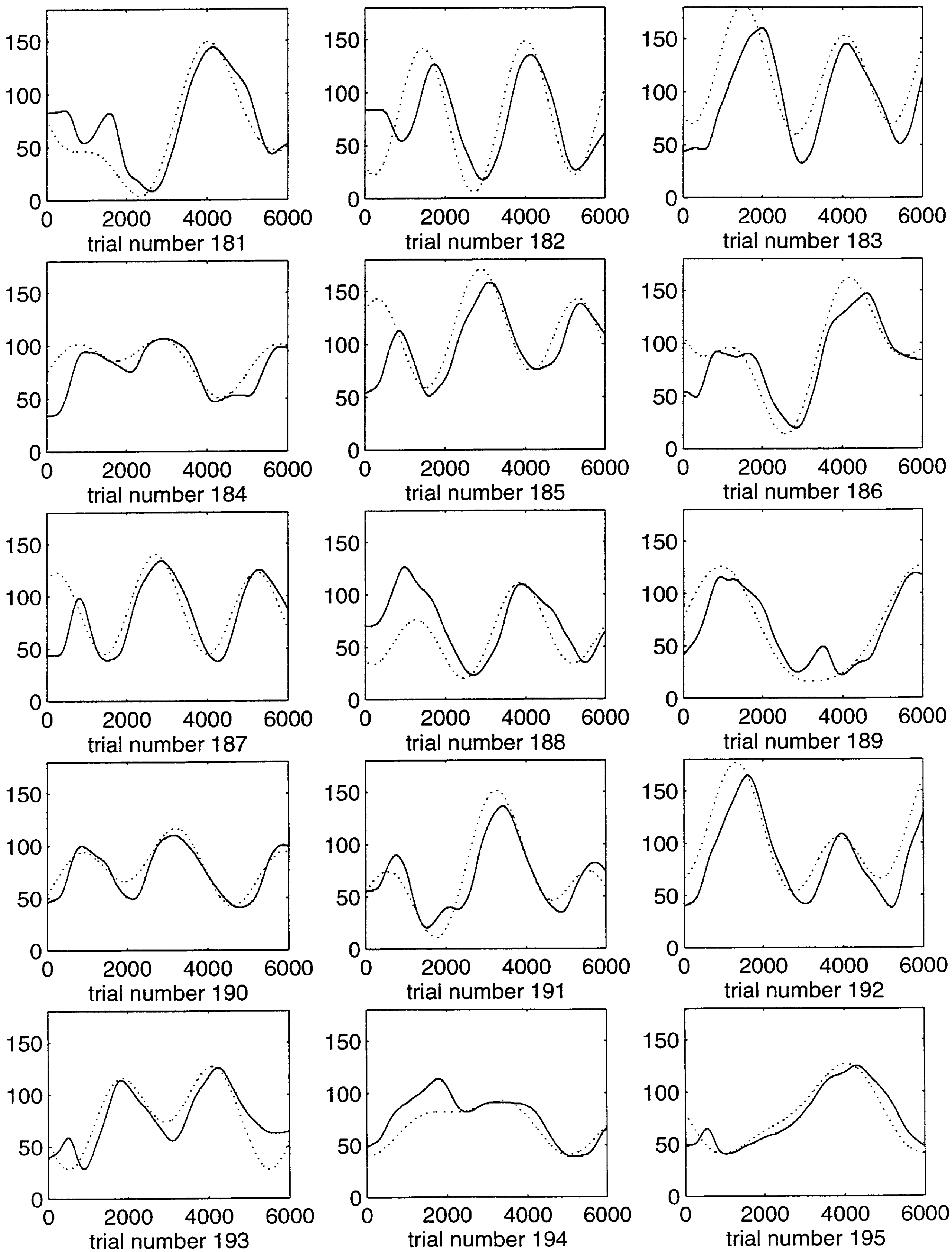
Data Plots: Post–Perturbation Trials



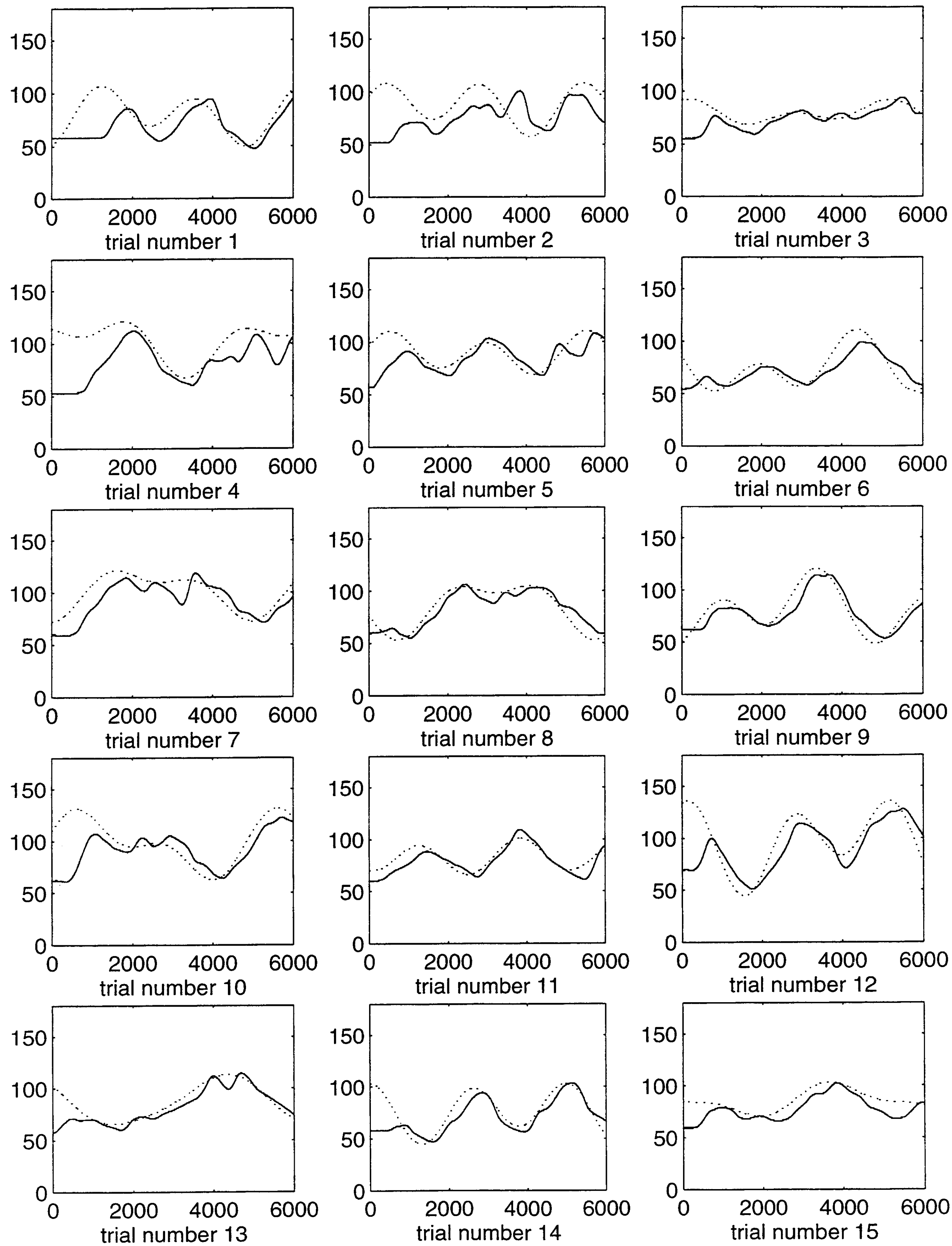
Data Plots: Post–Perturbation Trials

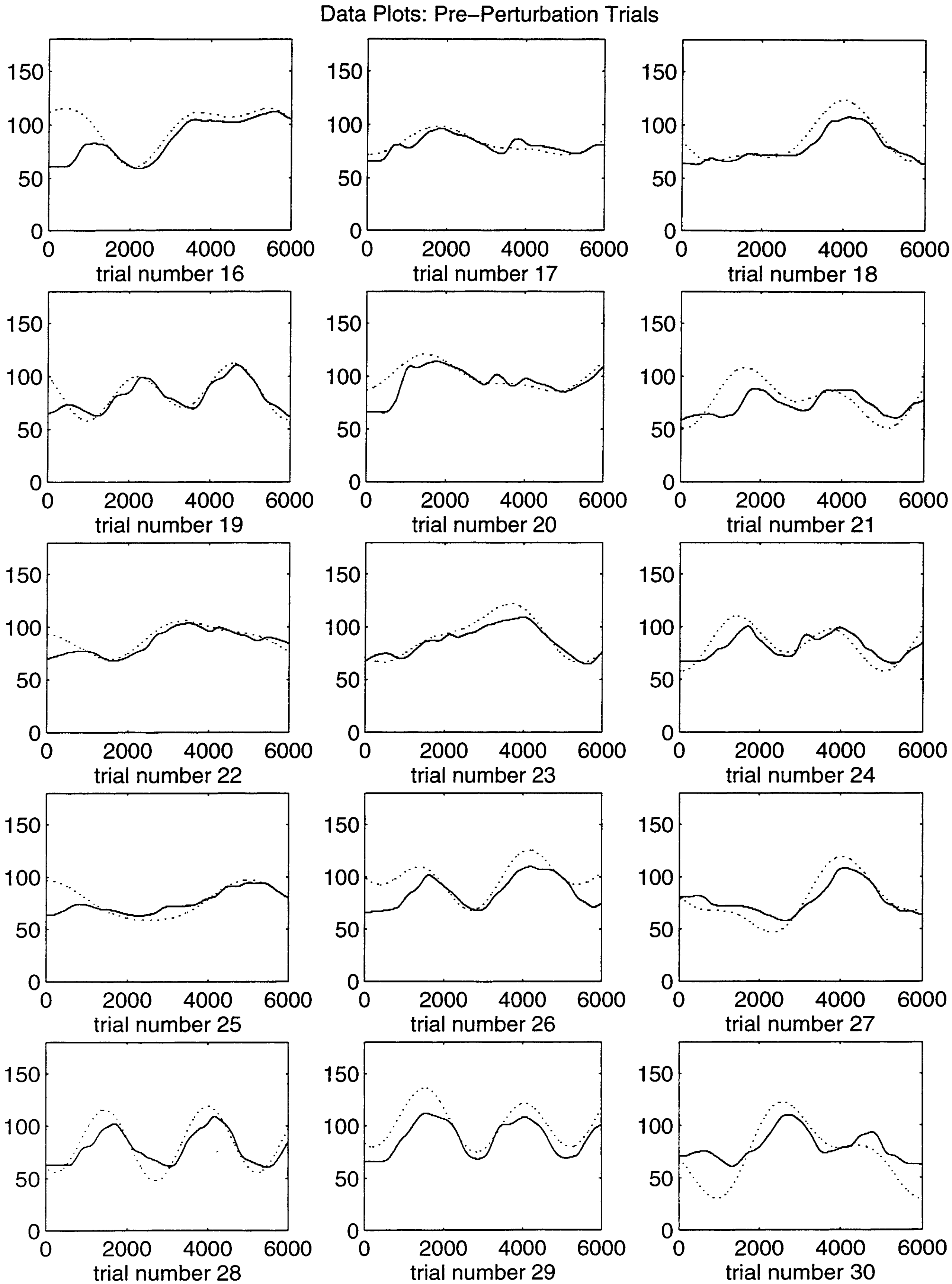


Appendix C

Data from the Amputee Subject

Data Plots: Pre-Perturbation Trials




Data Plots: Pre-Perturbation Trials
150
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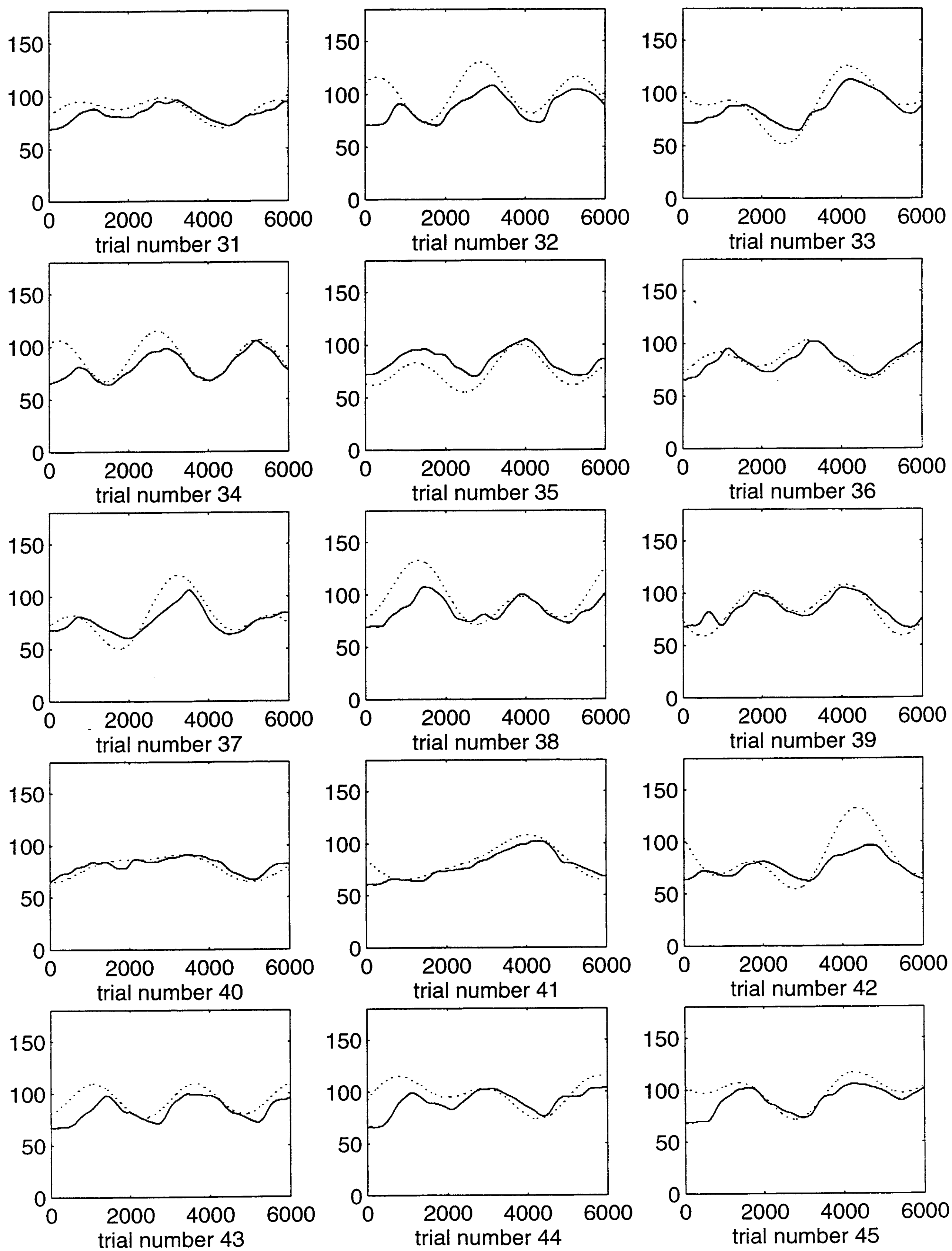
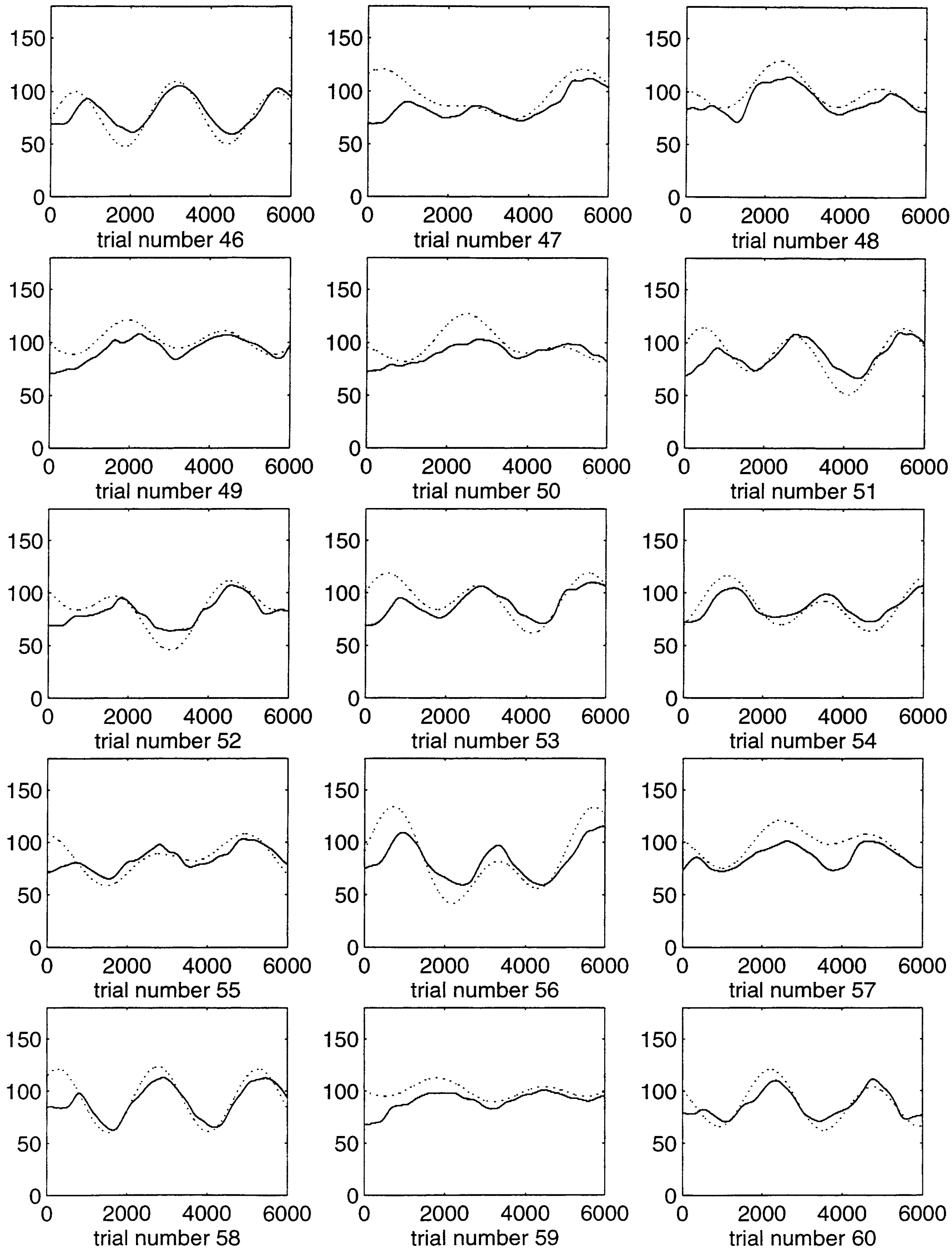
50
0
0
2000
4000
6000
trial number 16
trial number 17
trial number 18
trial number 19
trial number 20
trial number 21
trial number 22
trial number 23
trial number 24
trial number 25
trial number 26
trial number 27
trial number 28
trial number 29
trial number 30

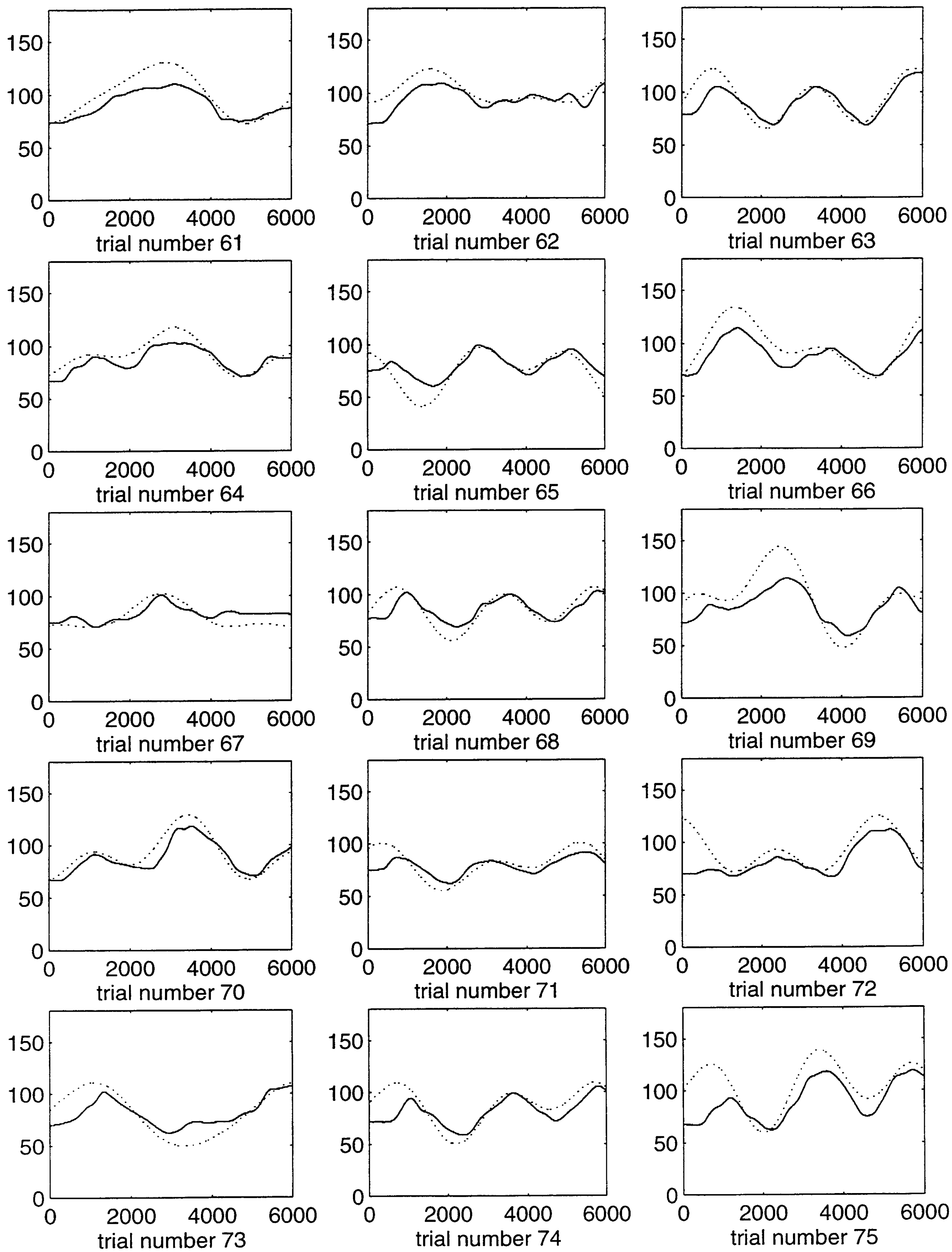
Data Plots: Pre–Perturbation Trials

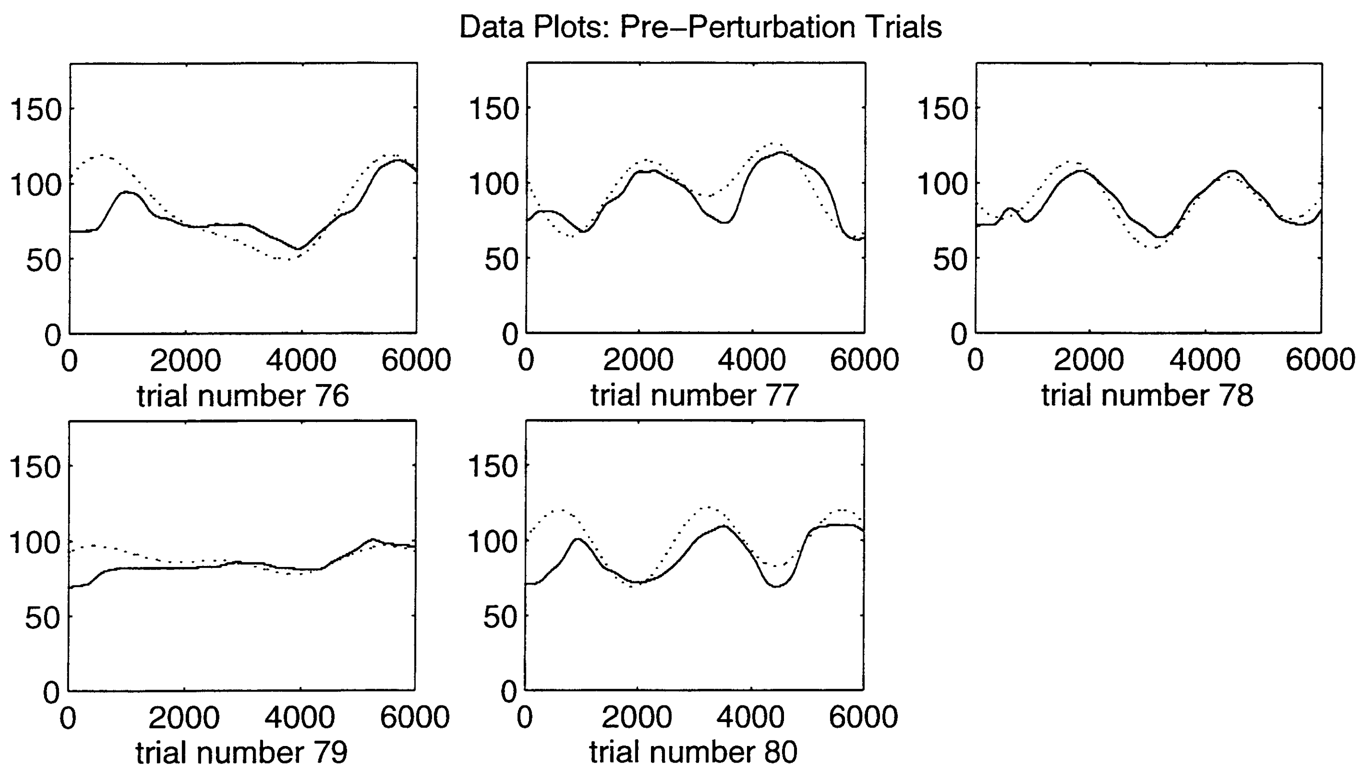


Data Plots: Pre-Perturbation Trials



Data Plots: Pre–Perturbation Trials

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50
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0
2000
4000
6000
trial number 76
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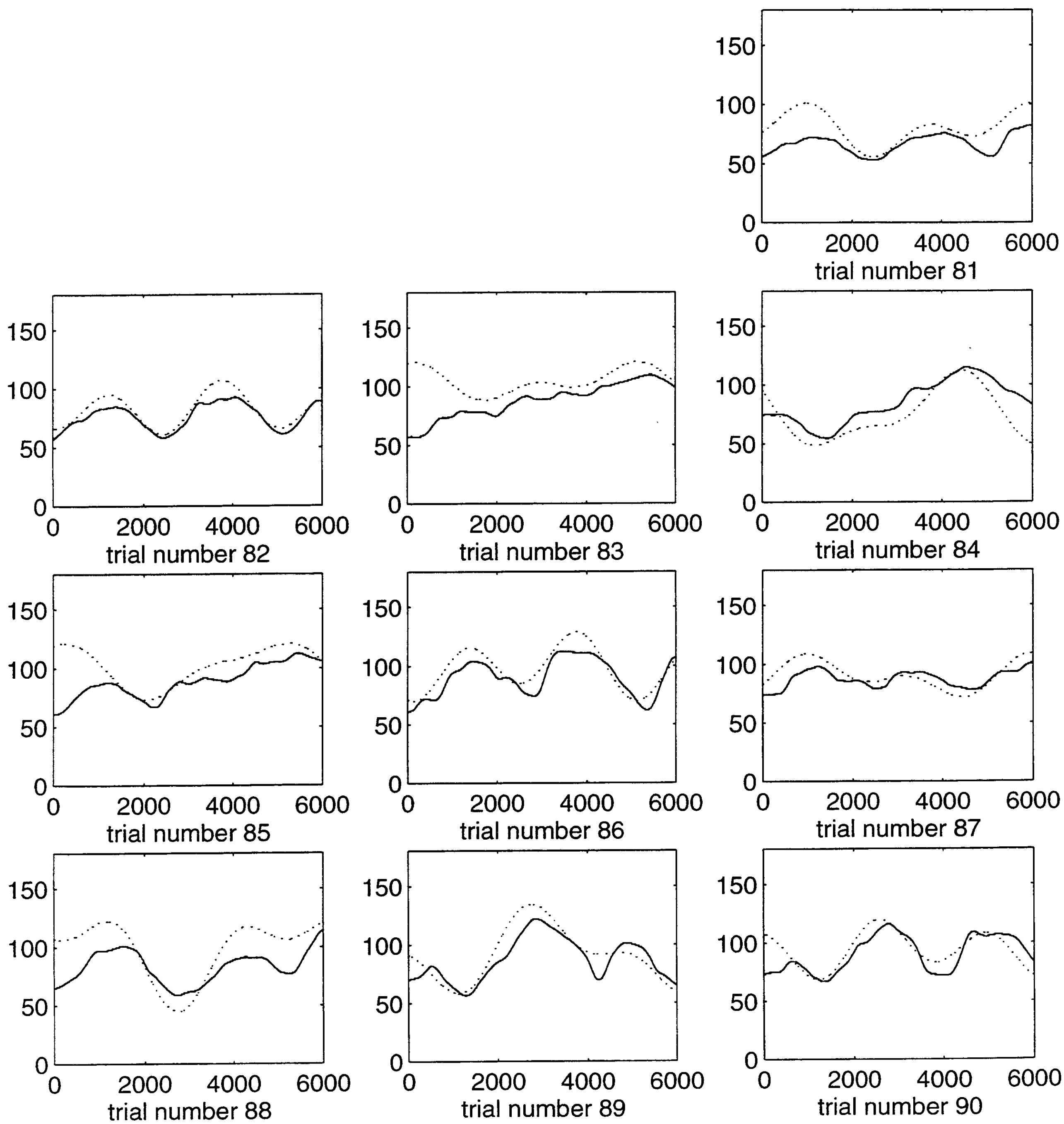
trial number 79
trial number 80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trial number 86
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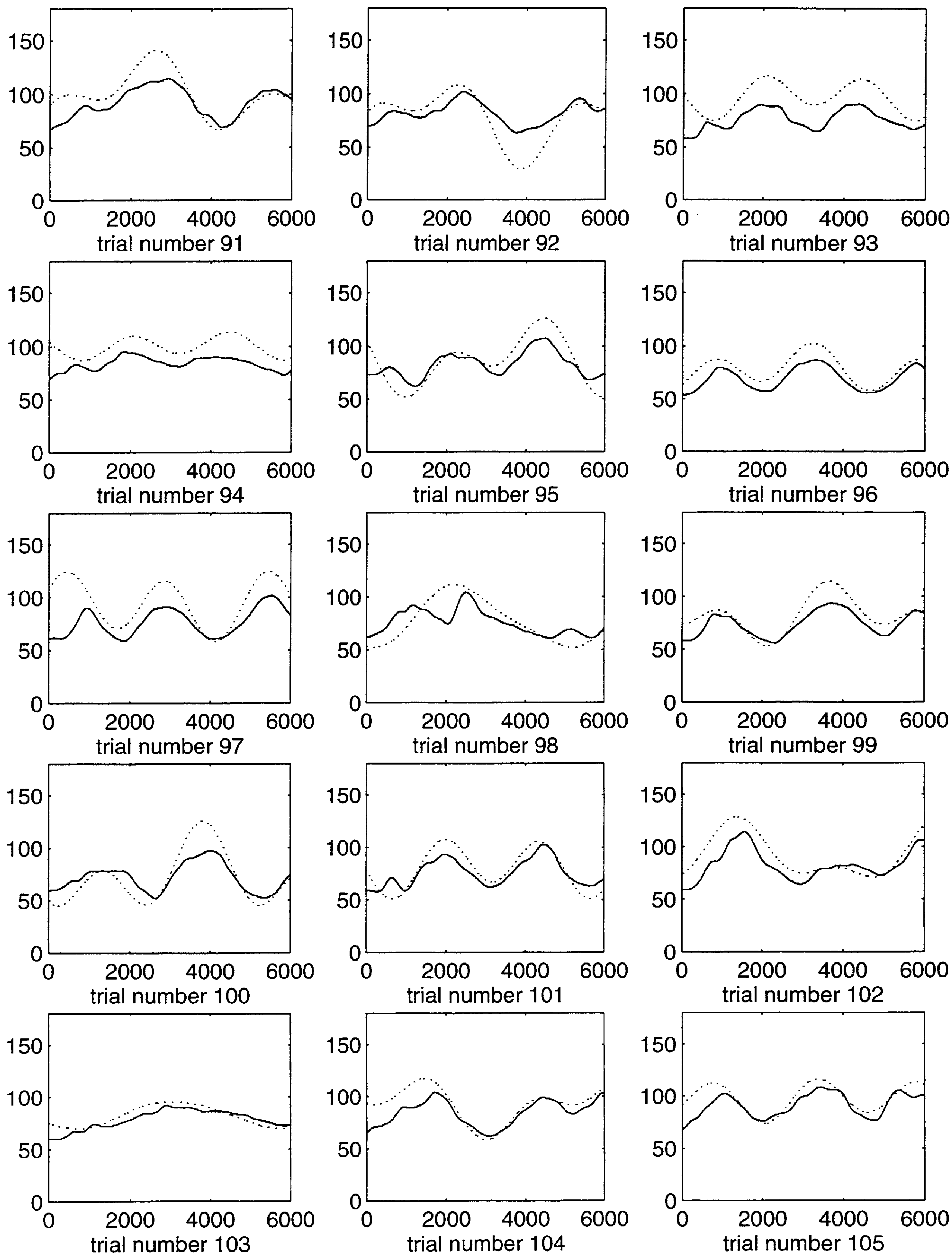
trial number 89
trial number 90

Data Plots: Perturbation Trials



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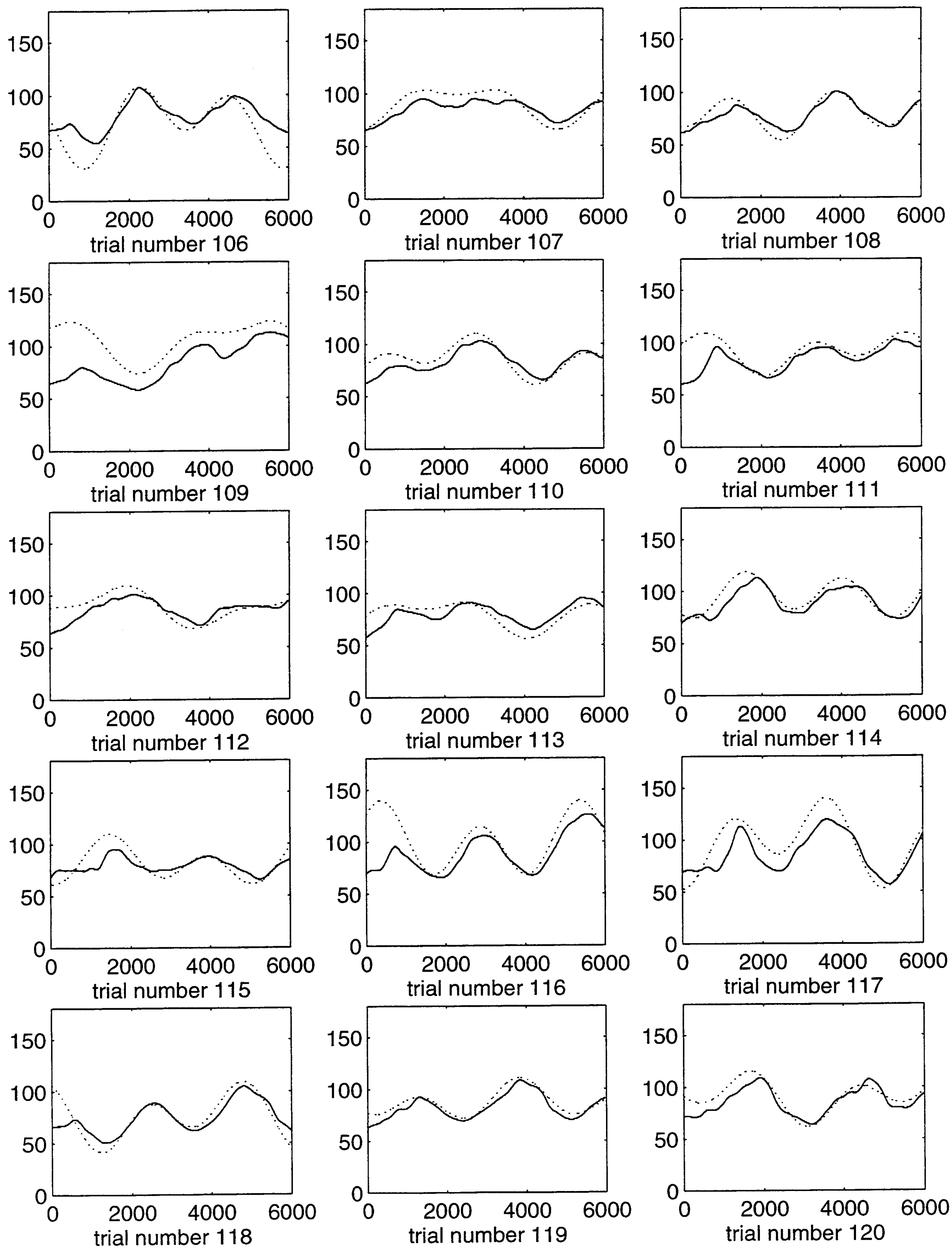

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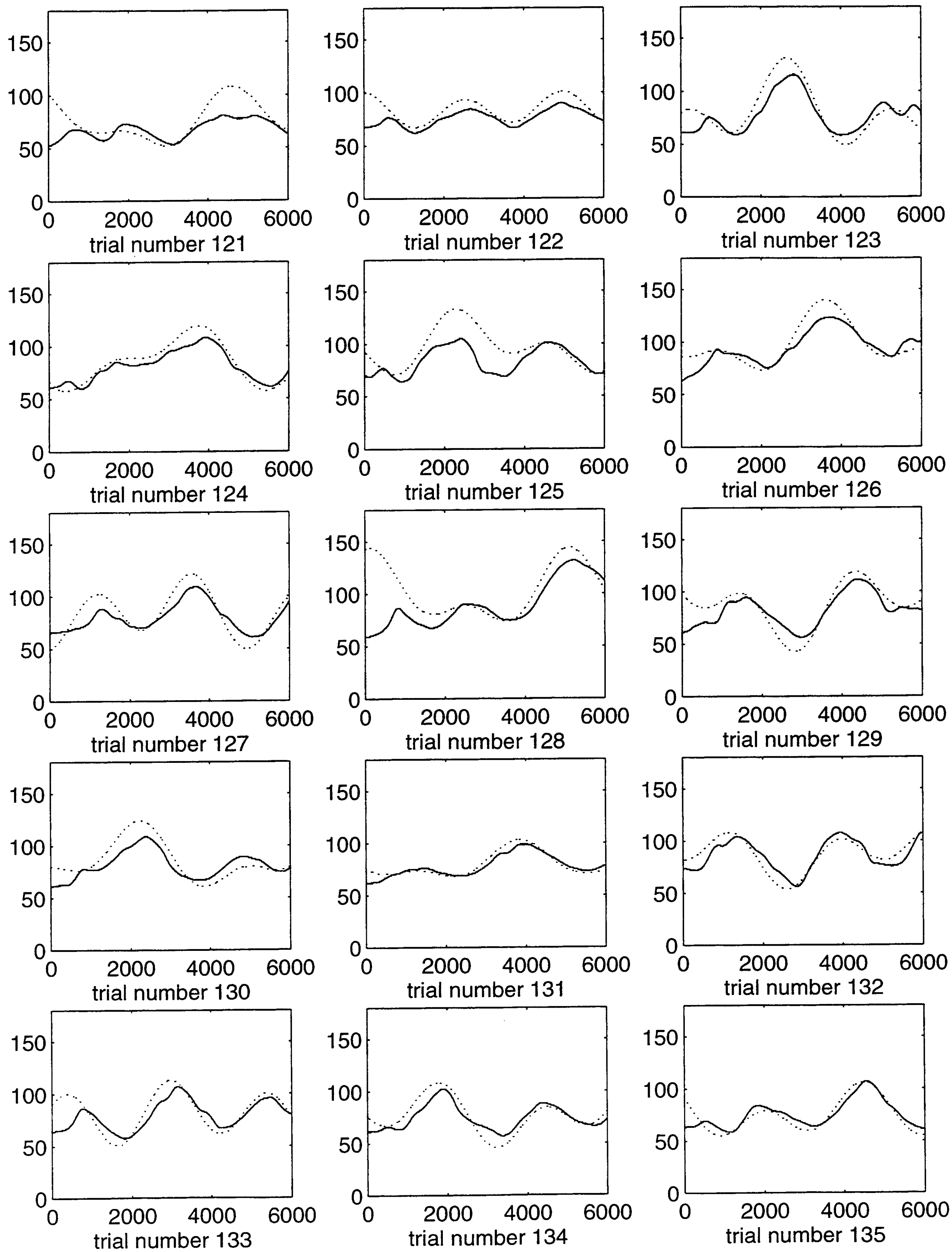

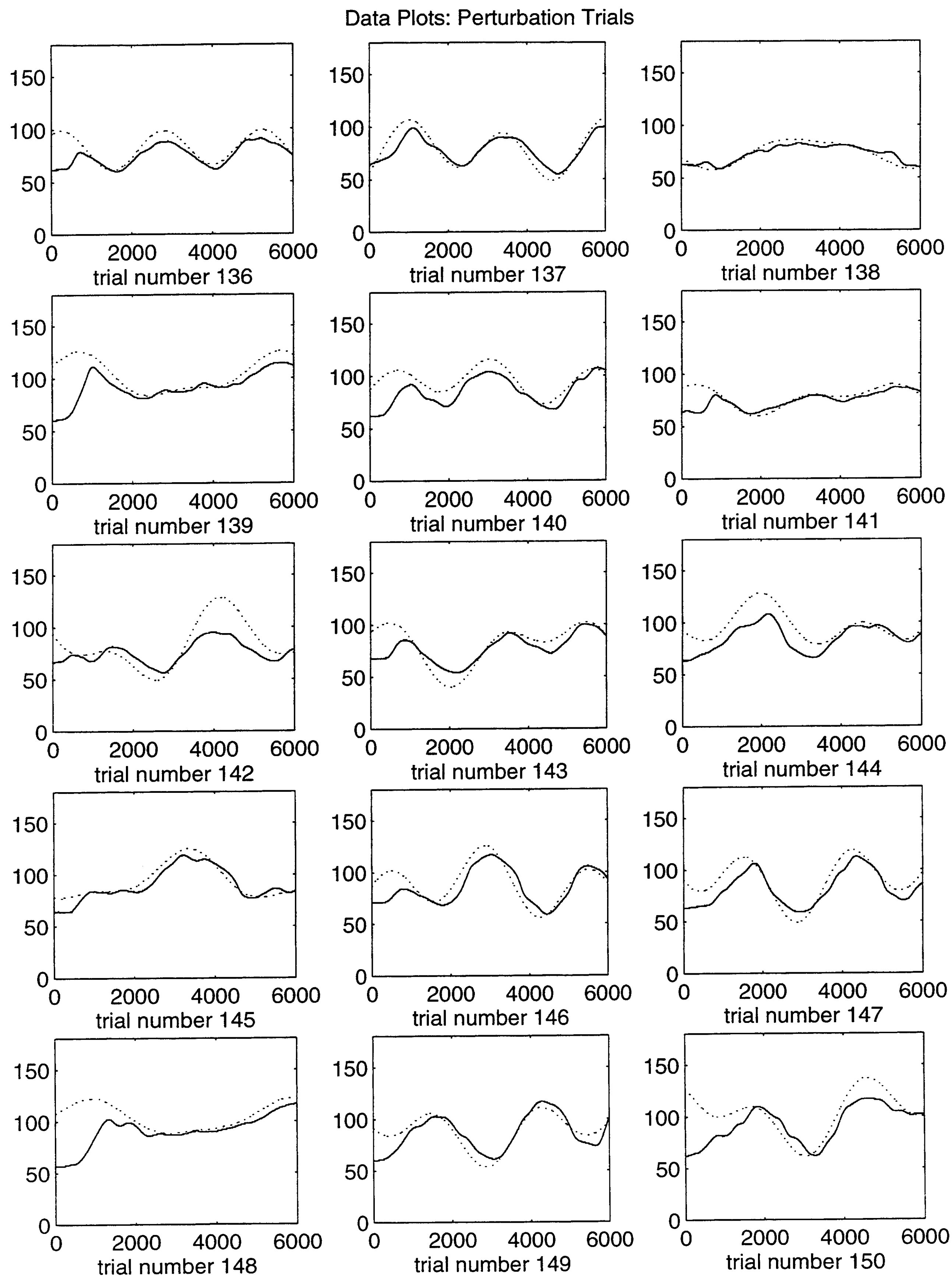

Data Plots: Perturbation Trials
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0
2000
4000
6000
trial number 136
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trial number 138
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trial number 140
trial number 141
trial number 142
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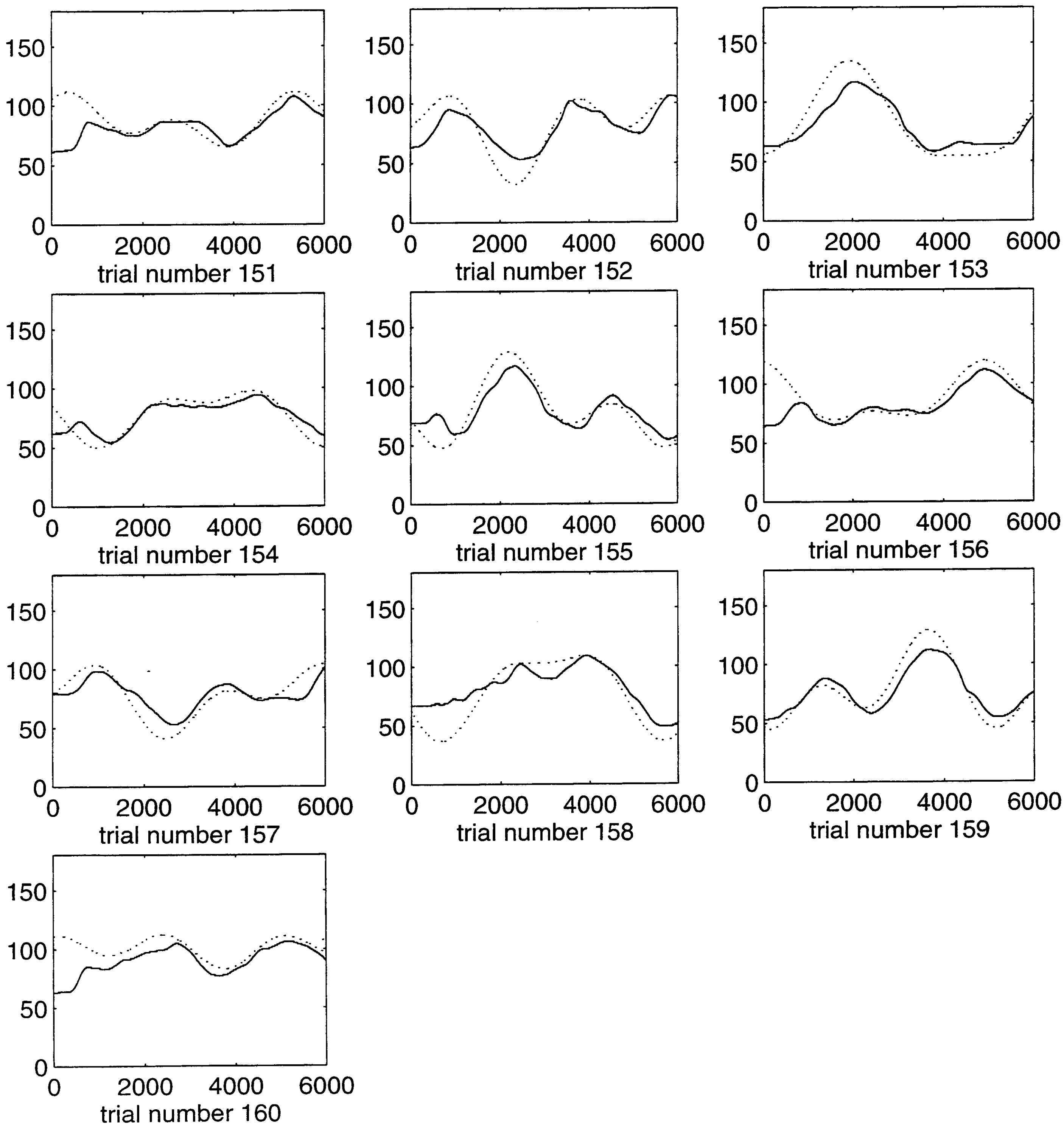
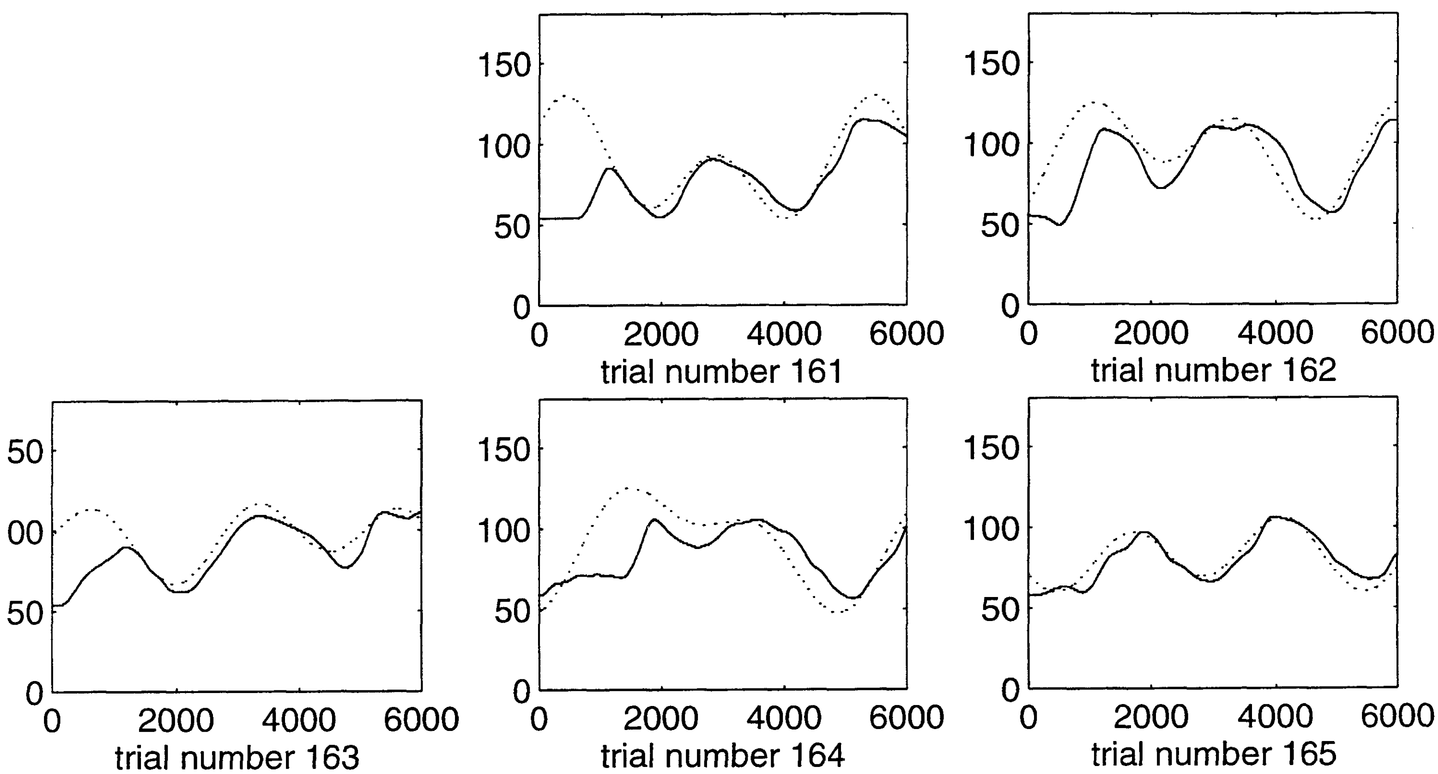
trial number 145
trial number 146
trial number 147
trial number 148
trial number 149
trial number 150

Data Plots: Perturbation Trials




150
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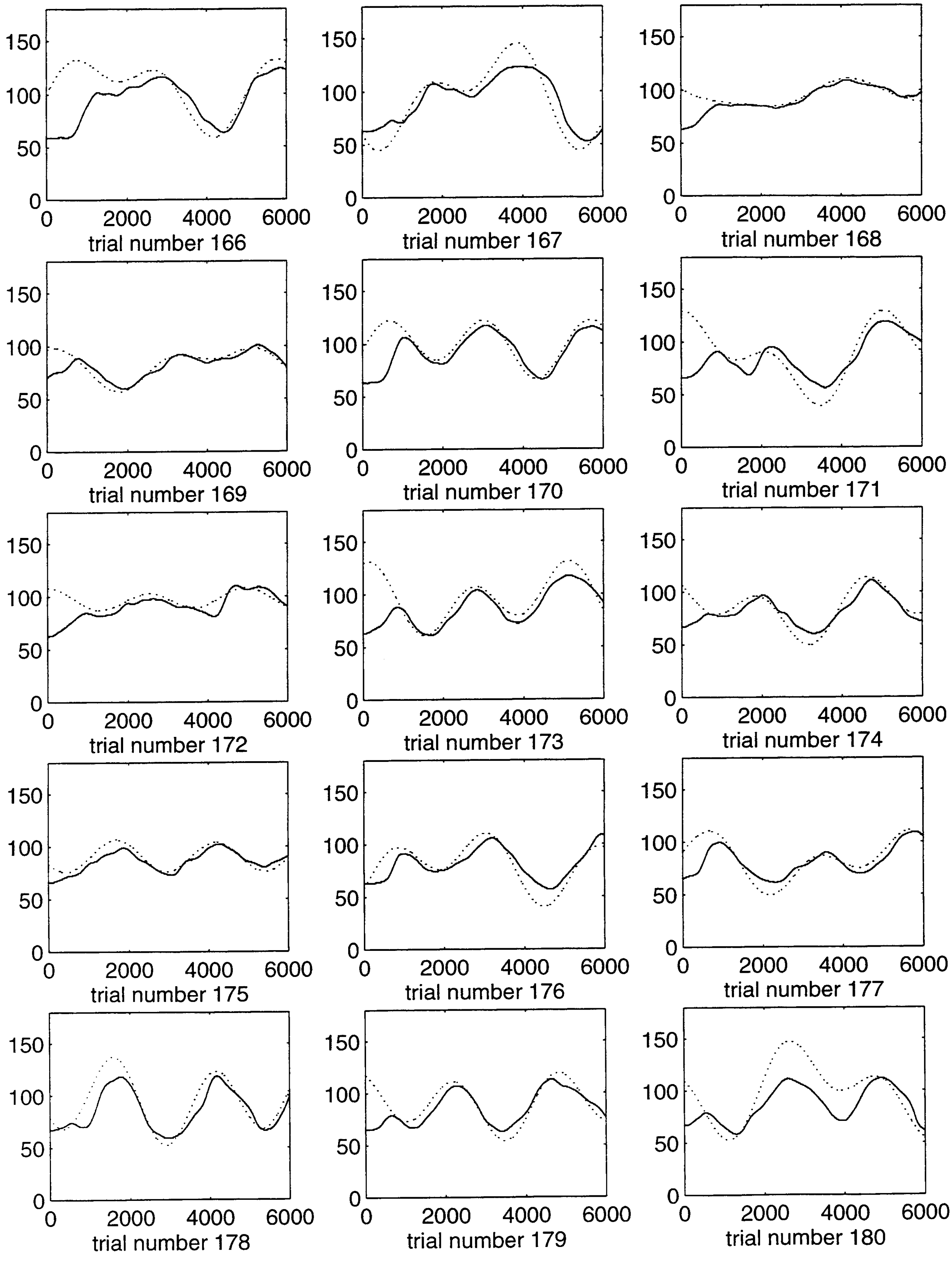
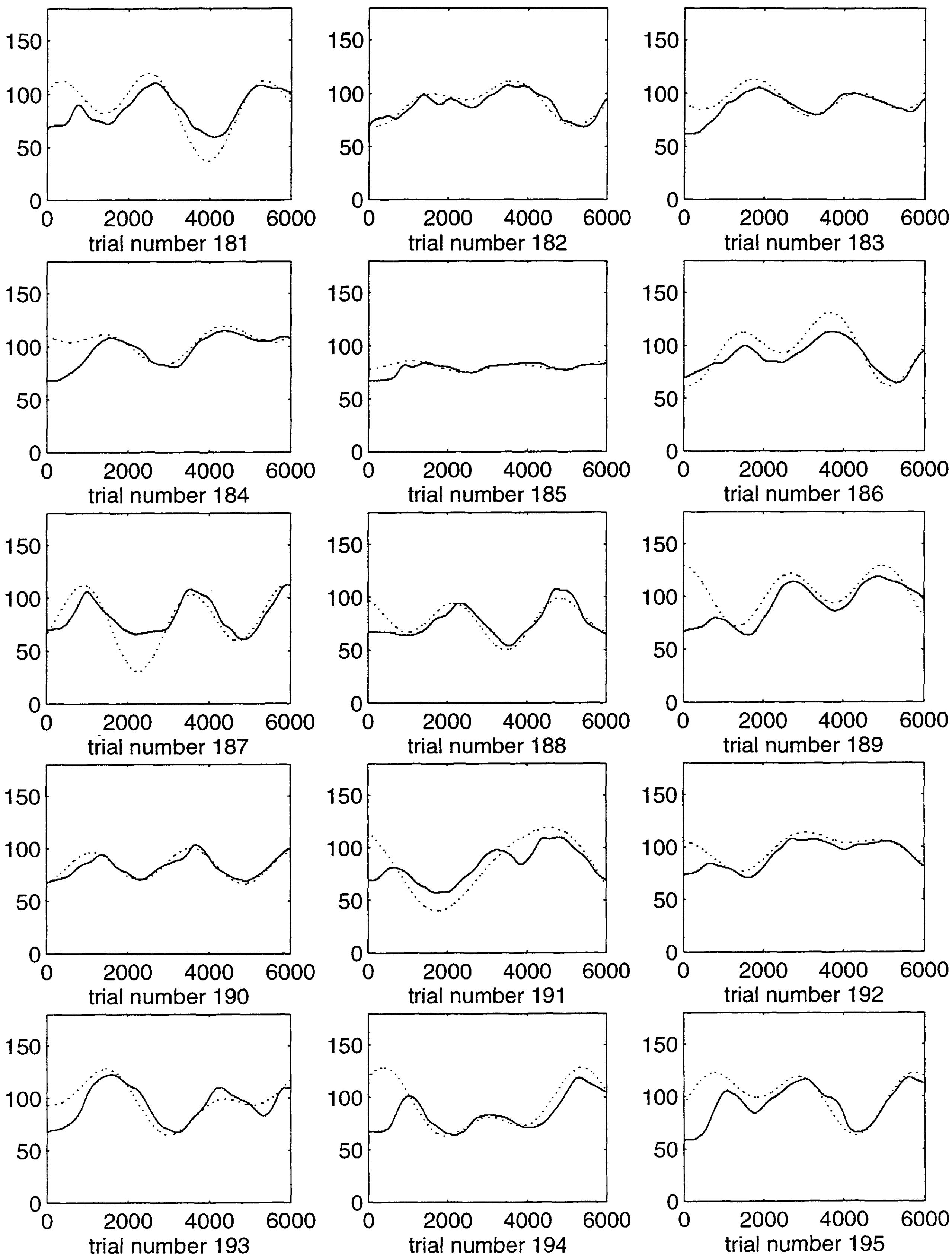
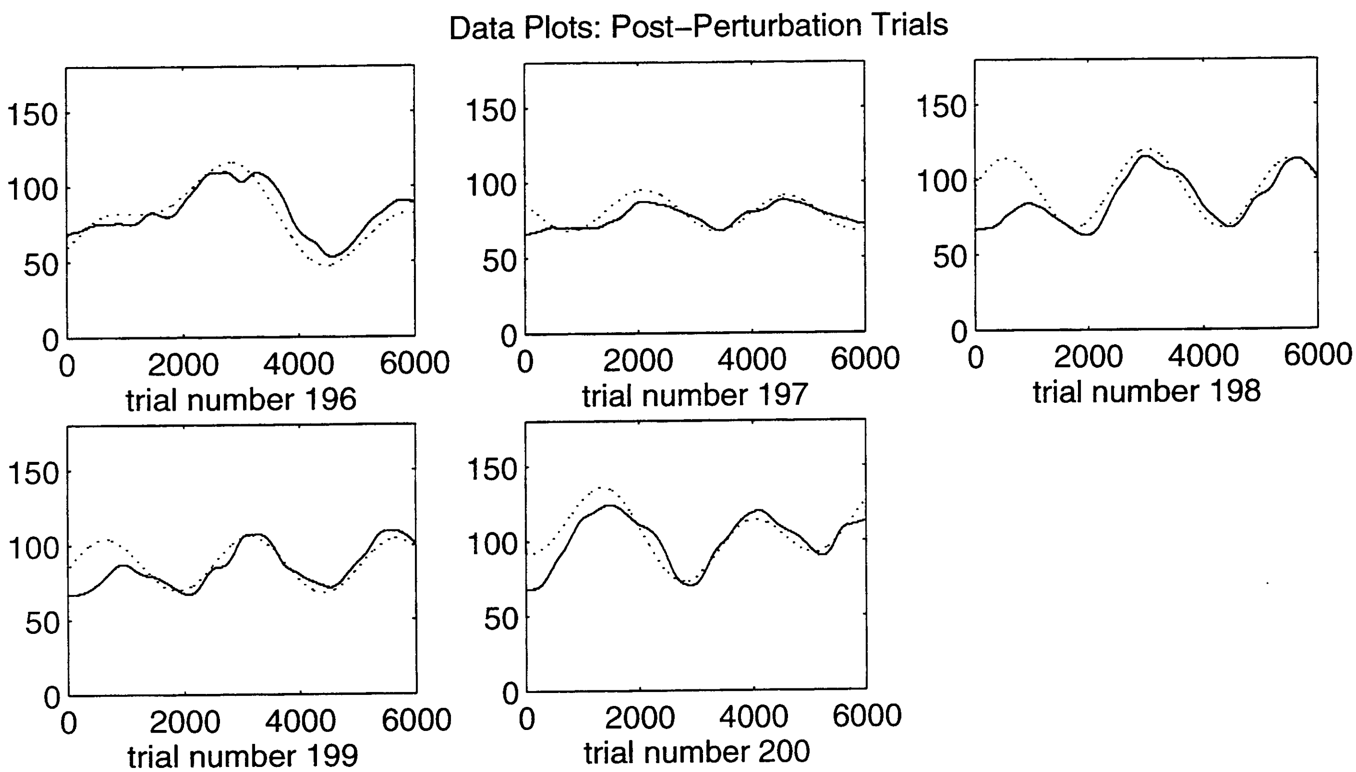
50
0
0
2000
4000
6000
trial number 161
trial number 162
trial number 163
trial number 164
trial number 165

Data Plots: Post–Perturbation Trials



Data Plots: Post–Perturbation Trials




Data Plots: Post–Perturbation Trials
150
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2000
4000
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trial number 196
trial number 197
trial number 198
trial number 199
trial number 200

Appendix D

Informed Consent Document

Informed Consent Document

Title of Study:	Investigation of Amputee and Non-amputee Execution of a Constrained-Motion Task
Principal Investigator:	Prof. Neville Hogan
Associated Investigator:	Brandon Rohrer

Purpose of Study

The long term goal of this research is to improve prosthesis design through a better understanding of how people use tools and interact with the physical world. The particular research described here has two main goals: to find a means by which the performance of two prostheses can be objectively compared and to study how people turn a crank. Being able to objectively compare prostheses allows prosthesis designers to systematically improve them, and understanding how humans turn a crank may give deeper insight into how we use tools in general.

Experimental Protocol

Subjects will be seated and asked to rotate a crank in a vertical plane. Details of how subjects move and the forces they exert on the crank will be recorded. Analysis of how subjects respond to different loads on the crank will provide a measure of how different arm amputation prosthesis designs facilitate (or impede) adaptation to load variations encountered in activities of daily living.

Amputee subjects will wear and operate an artificial arm with a computer-controlled motor-driven elbow. This apparatus is similar to the arm prostheses worn by many amputees except that it may be programmed to emulate the behavior of any existing arm prosthesis design (or any future design). It is attached to a socket that an amputee can wear on the amputated limb, supported by a standard commercial harness. A standard split-hook terminal device is used.

In these experiments the emulator will be programmed to emulate a common type of upper-arm prosthesis in which elbow flexion occurs when the amputee pulls on a cable by rounding the shoulders. The computer will sense how hard the amputee is pulling on the cable and the motor in the elbow will be used to help the amputee to move the elbow somewhat the way power steering helps a driver steer an automobile.

The crank is not motor-driven but weights and frictional loads may be applied to change the force required to move it. The forces required will be within the range encountered in activities of daily living.

Subjects will be seated near the crank which will rotate in a saggital plane to one side of the subject. The chair will be adjusted so that subjects can turn the crank from top to bottom while sitting comfortably upright with the shoulders resting against the chair back. A padded shoulder strap will be used to minimize shoulder movement. For non-amputee subjects, a commercially-available wrist splint will be used to minimize wrist movement.

Amputee subjects:

Amputee subjects will view a computer screen displaying two cursors, one indicating the crank's position, the other a target position that will fluctuate randomly at frequencies within the normal range of voluntary movement. Subjects will be instructed to move the crank to follow the target as closely as possible for approximately 20 seconds.

Three groups of 80 trials each will be performed, the first and third with no load applied to the crank, the second with loads applied to the crank as described above. To avoid fatigue, subjects will be asked to rest briefly (for about half a minute) every 10 trials. We anticipate that this experiment will require a single session of two hours duration and no more than three such sessions.

Non-amputee subjects:

Non amputee subjects will be presented with visual cues to mark the top and bottom positions on the crank. Subjects will be asked to move the crank back and forth between the two positions at a constant cadence. Twelve groups of five trials will be performed, respectively (a) with and without pauses at each extreme position, (b) with and without loads applied to the crank and (c) at three different speeds ("comfortable speed", "rapid" or slow"). To move "rapidly", the experimenter will verbally instruct the subject to adjust speed until a single upward or downward movement lasts approximately three-quarters of a second. To move "slowly", the experimenter will verbally instruct the subject to adjust speed until a single upward or downward movement lasts approximately 6 seconds.

Each trial will last for between 30 and 60 seconds and subjects will be asked to rest briefly (for about half a minute) between trials. We anticipate that this experiment will require a single session of two hours duration and no more than three such sessions.

Risks and Benefits

The crank is an un-powered device and subjects will be seated such that there is no danger of the crank handle striking them as it turns. Electronic instrumentation on the crank to measure force and motion operates at low voltage and current from a fully grounded electrical supply. Its use involves no risk.

The actuator and sensors of the prosthesis emulator are also powered from a fully grounded electrical supply. Because the emulator has a motor, it is in principle capable of striking the wearer should it malfunction. Several levels of protection have been provided to prevent this from happening. The computer monitors the elbow speed and will automatically shut off the system if the elbow moves too rapidly. In addition, four separate human-operated "kill switches" (at least one of which is accessible to the subject and at least one of which is accessible to the experimenter) are provided. Operation of any of these switches will shut off the system whether the computer is functioning or not. To shut off the system, electrical power is immediately removed from the motor. At the same time a braking force is applied to bring the elbow to rest without relying on the functioning of any electronic components. The principal investigator has over a decade of experience using this apparatus for prosthesis research at MIT and has maintained a 100% safety record.

There are no known benefits to the subjects participating in this study.

Payment

Amputee subjects will be paid $50 as a partial compensation for their time spent in the lab. Subjects will also be refunded their travel expenses, if any. Non-amputee subjects will not be compensated.

Subject Agreement

I volunteer for this project and am free at any time to seek further information regarding the experiment. In addition, I am also free to withdraw consent and discontinue participation at any time.

I understand that any monetary compensation I am to receive for participating in this study will be pro-rated if I choose to withdraw before completion of the experiment.

I understand that my name will be withheld from any publication resulting from these experiments.

In the unlikely event of physical injury resulting from participation in this research, I understand that medical treatment will be available from the M.I.T Medical Department, including first-aid emergency treatment and follow-up care as needed, and that my insurance carrier may be billed for the cost of such treatment. However, no compensation can be provided for medical care apart from the foregoing. I further understand that making such medical treatment available; or providing it, does not imply that such injury is the Investigator's fault. I also understand that by my participation in this study, I am not waiving any of my legal rights. Further information can be obtained by calling the Institutes Insurance and Legal Affairs Office at 253-2822.

I understand that I may also contact the Chairman of the Committee on the Use of Humans as Experimental Subjects, M.I.T. 253-6787, if I feel I have been treated unfairly as a subject.

I have read the above consent document and understand the experiments described in the document. I agree to participate in the experiments as a subject.

The project investigators retain the right to cancel or postpone the experimental procedures at any time they see fit.

Subject's Signature:________________________________ Date:____________

Subject's Name:___________________________________

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